Université du Québec

INRS
Institut national de la recherche scientifique

Énergie Matériaux Télécommunications

RADIO FREQUENCY MICROSTRIP LINE MODEL OF CIRCULATORY SYSTEM VESSELS FOR CARDIOVASCULAR APPLICATION

Par

Sada D Saleh El Dersse

Mémoire ou thèse présenté(e) pour l'obtention du grade de

Maître ès Sciences (M.Sc.) en sciences de

[Master's in Telecommunications]

Jury d'évaluation

Président du jury et examinateur interne	Tatu, Serioja Ovidiu Énergie Matériaux Télécommunications INRS
Examinateur externe	Mojabi, Puyan Electrical and Computer Engineering University of Manitoba
Directeur de recherche	Tarek Djerafi Énergie Matériaux Télécommunications INRS

© Droits réservés de « Sada El Dersse », 10, 2019

REMERCIEMENTS

I would like to express all my thanks to my supervisor Prof Tarek Djerafi for giving me the opportunity to work in his team and guiding me through my MASc program. All my gratitude to his patience, and availability and his great interest in this work.

Special thanks to the external and internal jury members who have kindly accepted to assess this work.

Finally, my deepest gratitude to my husband, and family for their unconditional support and love.

RÉSUMÉ

Toutes les sept minutes, une personne décède d'une maladie cardiaque ou d'un accident vasculaire cérébral au Canada. En outre, selon l'organisation canadienne de cardiologie, au moins 1 000 patients sont nés avec un seul ventricule fonctionnel, qui nécessite plusieurs chirurgies à cœur ouvert pour rendre la circulation durable. L'opération Fontan permet aux enfants nés avec ce type de cardiopathie congénitale de survivre, mais en raison de l'absence de deuxième ventricule, le sang s'écoulant dans les poumons. En raison de la difficulté d'une étude réaliste de l'évolution du système circulatoire dans le corps humain, moins de 50% des bébés survivent à l'âge adulte en raison de maladies chroniques. D'où l'importance que les modèles permettant de décrire le système cardiovasculaire fournissent des données plus précises pour déterminer l'évolution de conditions du patient.

Certains de ces modèles dépendent de l'ajustement ou de l'estimation de la courbe ; les autres utilisent les équations de fluide. Dans l'approche proposée, l'analogie entre le système cardiovasculaire et la théorie de la ligne de transmission, qui tire son origine de l'équivalence d'équations de Navier et de Maxwell, est utilisée. L'utilisation du logiciel ADS pour simuler ce modèle offre une flexibilité dans la conception en raison de ses fonctionnalités telles que l'ajustement qui permet de modifier une ou plusieurs valeurs de paramètre et d'illustrer l'effet sur la sortie sans simuler à nouveau le système.

L'objectif de cette étude est de valider un modèle miniaturisé du système circulatoire humain. Les résultats démontrent que l'augmentation de la fréquence et la variation de la longueur n'affectent pas la forme d'onde du pouls, ce qui a conduit notre approche à la conception d'un petit modèle utilisant une ligne de transmission à microruban et à l'obtention des mêmes résultats.

La ligne de transmission microruban a été mesurée sur la base de certains paramètres de vaisseaux sains. Ce Modèle permettra d'appliquer les caractéristiques de la ligne de transmission sur le corps humain afin de définir certains paramètres et d'obtenir un meilleur aperçu de ses aspects physiques et l'étude des effets pathologiques.

Mots-clés : Advanced design system, microruban, modèle à éléments localisés, système cardiovasculaire, système circulatoire, Vaisseaux sanguins.

ABSTRACT

Every 7 minutes in Canada someone dies from heart disease or stroke. One of these causes of cardiovascular disease is atherosclerosis-the narrowing and eventual blockage of arteries by the deposition of fatty plaques on the walls of the artery. In addition, there is at least 1,000 patients born with only one functional ventricle according to the Canadian heart organization, which need multiple open-heart surgeries to make the circulation sustainable. The Fontan operation permits children born with this type of congenital heart disease to survive but due to the lack of a second ventricle pumping blood to the lungs less than 50% of the babies survive to adulthood due to chronic illnesses. Due to the difficulty of realistic study of the changing on the circulatory system in the human body, models that can describe the cardiovascular system is important to offer closer data to define the patient conditions. Some of these models are depending on curve fitting or estimation; the others are followed some fluid equations.

The objective of this study is to validate a miniaturized model of the human circulatory system. The research depends on a study that has been done to certify our concern by demonstrating the increase in the frequency and changing in the length does not affect on pulse waveform that leads to our approach to design a small model using a microstrip transmission line and accomplishment the same results. In the proposed approach, the analogy between the cardiovascular system and transmission line theory, which takes source from Navier stokes and Maxwell equations equivalence is used. The use of ADS software to simulate this model provides flexibility in the design because of the features of this software such as the tuning, which enable the modification of one or more parameter value and illustrate the effect on the output without re-simulating the entire design. The rules that have been used to reach the equivalent microstrip substrate are studied and presented. The equitable procedure has been done to achieve reasonable permittivity and thickness.

Microstrip transmission line has been measured based on some healthy vessels parameters that leads to a new novelty, which may enable to apply the characteristics of the transmission line on the human body to define some parameters and obtaining a better insight into its physical aspects at the same time studying the effects of pathological conditions.

Keywords: Advanced design system, blood vessels, circulatory system, cardiovascular system, lumped element model, microstrip.

SOMMAIRE RÉCAPITULATIF

0.1 Système de circulation sanguine

Le système de circulation est l'un des systèmes les plus compliqués du corps humain, car il est responsable de la délivrance du sang, des nutriments, de l'oxygène et d'autres hormones à partir des cellules du corps. Il se compose de trois parties : cardiovasculaire (le cœur), pulmonaire (pulmonaire), systémique (artères, veines et coronaires) qui fonctionne conjointement de manière organisée et systémique.

La partie cardiovasculaire implique le cœur, les vaisseaux sanguins et le sang.

1.1.1 Le cœur

Le cœur est situé légèrement à gauche de la poitrine et les parois du cœur sont constituées d'un type unique de muscle appelé muscle cardiaque. Les cellules musculaires cardiaques sont disposées dans un réseau qui permet au cœur de se contracter et de se détendre rythmiquement et involontairement sans se fatiguer. Le cœur comporte quatre cavités(Figure 0-1), deux à gauche et deux à droite. Les cavités du haut sont appelées atrium et les cavités du bas, appelées ventricules. Les oreillettes se remplissent de sang provenant du corps ou des poumons. Les ventricules sont responsables de pomper le sang vers le corps après avoir reçu le sang des oreillettes. Les oreillettes et le ventricule sont séparés par la valve auriculo-ventriculaire (A-V), (valve bicuspide) du côté gauche et (valve tricuspide) du côté droit.

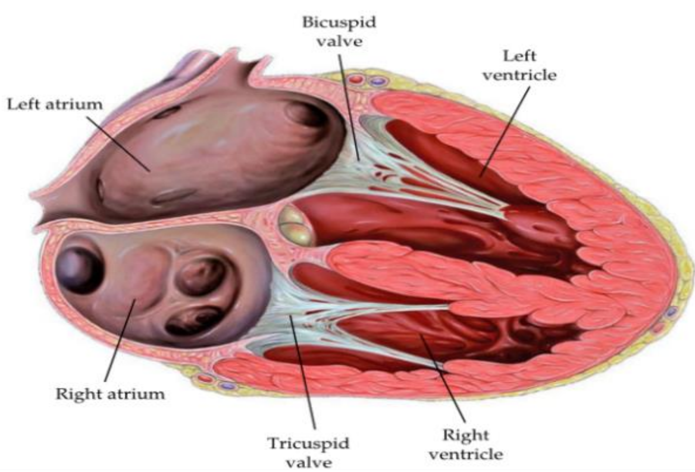


Figure 0-1: Une illustration d'un cœur sectionné [6].

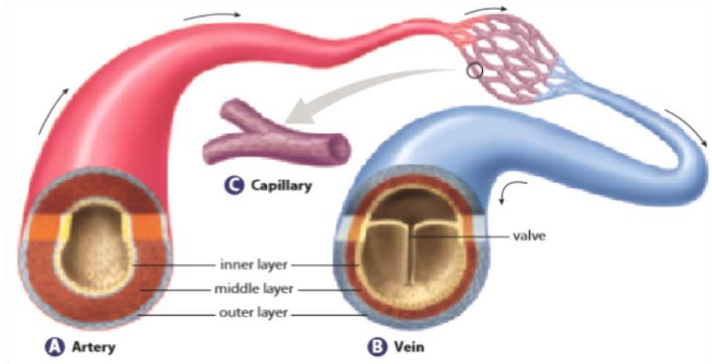


Figure 0-2: Artères (A) veines (B) Constitués de trois couches. (C) Les capillaires sont constitués d'une seule couche de cellules

0.2 Vaisseau sanguin

Il existe trois types de vaisseau illustrés dans la (Figure 0-2) et chaque vaisseau a une fonction qui dépend de sa position et de ses propriétés.

Artères : Les artères transportent le sang du cœur au corps. Ce type de vaisseau a une paroi élastique qui s'étend en fonction du flux sanguin lors de la contraction des ventricules et qui se compresse lorsque les ventricules se détendent. L'aorte (côté gauche) et l'artère pulmonaire (côté droit) sont les principales artères qui quittent le cœur. L'aorte est responsable de faire passer le sang oxygéné du côté gauche du cœur au reste du corps, et l'artère pulmonaire accomplis l'opération inverse qui fournit le sang désoxygéné (pauvre en oxygène) aux poumons du côté droit du corps.

Veines : sont les transporteurs de sang vers le cœur. Leurs parois sont plus minces que les artères et ne sont pas aussi élastiques que les artères, ce qui n'aide en rien le mouvement du sang. Au lieu de cela, les veines ont des valves à sens unique qui empêchent le sang de refluer. Ces valves jouent un rôle important dans la partie inférieure du corps, car elles veillent à ce que le sang afflue vers votre cœur contre l'attraction gravitationnelle. La veine cave est la veine principale.

Capillaires : Les capillaires sont les plus petits vaisseaux sanguins et ont une paroi très mince pour permettre à l'oxygène et aux nutriments de passer à travers la cellule du corps et des cellules tissulaires jusqu'au sang. Un réseau de capillaires relie les artères et les artérioles aux sites et aux veines [5].

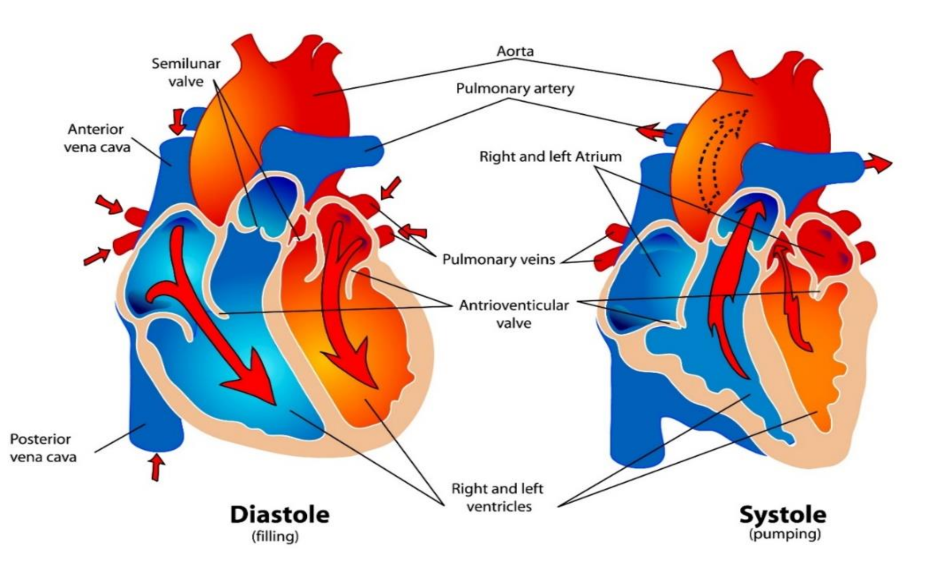


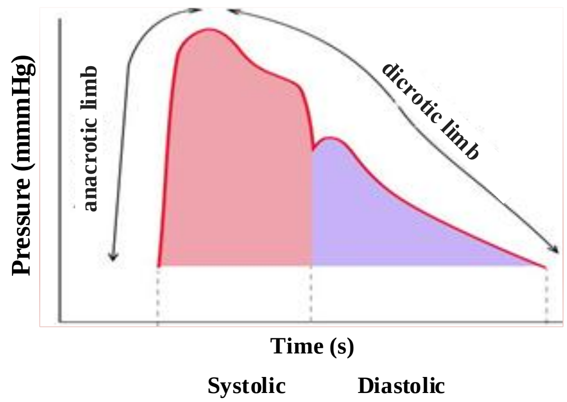
Figure 0-3: Phases du cycle cardiaque du chapitre 9: "Heart muscle: the heart as a pump and the function of heart valves"

0.3 Cycle Cardiaque

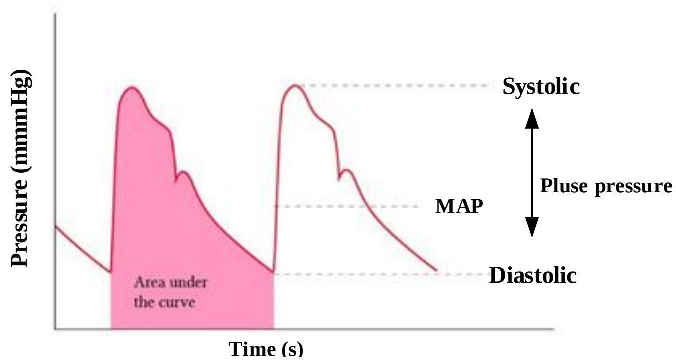
Le cycle cardiaque est la séquence d'un battement de cœur. Simplement, il s'agit de la contraction simultanée des deux oreillettes, suivie de la contraction simultanée des deux ventricules en une fraction de seconde. Le cœur est constitué de cellules musculaires cardiaques qui se connectent les unes aux autres. Ainsi, lorsqu'on se contracte, elles stimulent les cellules adjacentes et elles se contractent toutes. Le cœur se repose entre les battements. Il ne peut que respirer en aérobiose.

Le rythme cardiaque a deux phases : Systole : est le terme pour la contraction. Cela se produit lorsque les ventricules se contractent, en fermant les valves A-V et en ouvrant les valves semi-lunaires, comme indiqué à la (Figure 0-3), pour pomper le sang dans les deux principaux vaisseaux quittant le cœur. Diastole : le terme de relaxation. Cela se produit lorsque les ventricules sont relâchés, ce qui permet à la contre-pression du sang de fermer les valves semi-lunaires et d'ouvrir les valves A-V [5], la Figure 0-3 notifiant que les deux phases se produisent dans les deux côtés dans des directions différentes pour permettre la circulation du sang dans le cœur.

0.4 Forme d'onde de pression



(a)



(b)

Figure 0-4: (a) Forme d'onde de pression représente les phases systole et diastole, (b) Pression de pouls avec valeur moyenne

La forme d'onde de pression artérielle est un élément clé de l'état hémodynamique et un indicateur de la transfusion adéquate des organes et du débit tissulaire. Il représente l'impulsion de la contraction ventriculaire gauche. La forme d'onde des poules artériel peut être séparée en trois phases, comme indiqué à la (Figure 0-4) :

- La phase systolique, caractérisée par une augmentation rapide de la pression jusqu'à un pic, suivie d'un déclin rapide. Cette phase commence avec l'ouverture de la valve aortique et correspond à l'éjection ventriculaire gauche comme mentionné précédemment.
- L'encoche dicrote, c'est un mouvement ascendant secondaire dans la partie descendante d'un tracé d'impulsions correspondant à l'augmentation transitoire de la pression aortique à la fermeture de la valve aortique.

- La phase diastolique, qui représente l'écoulement de sang dans la circulation périphérique.

0.5 **Pression sanguine**

Selon « United Kingdom (UK) heart organization », la pression artérielle est définie comme la force permettant de pousser le sang pendant le vaisseau lorsque le cœur bat (ce qui prend environ 0,8 seconde). Idéalement, pour les personnes en bonne santé, la pression artérielle devrait être inférieure à 120 mmHg (120/80), le chiffre le plus élevé étant la pression artérielle systolique, qui correspond à la pression la plus élevée lorsque le cœur bat et pousse le sang dans tout le corps. La pression la plus basse lorsque le cœur se détend entre les battements est la pression artérielle diastolique.

Maintenir la tension artérielle dans la valeur normale est très important. Une pression artérielle plus élevée impose un stress supplémentaire aux artères et au cœur. Cette tension peut rendre les artères plus épaisses et moins souples, ce qui conduit à un rétrécissement et à un encrassement, puis à une crise cardiaque ou à un accident vasculaire cérébral. La (Figure 0-5) montre les plages de lectures de pression artérielle élevées, basses et saines. Par conséquent, la plupart des recherches sur la modélisation et la représentation considèrent ces valeurs pour évaluer la précision de leurs travaux.

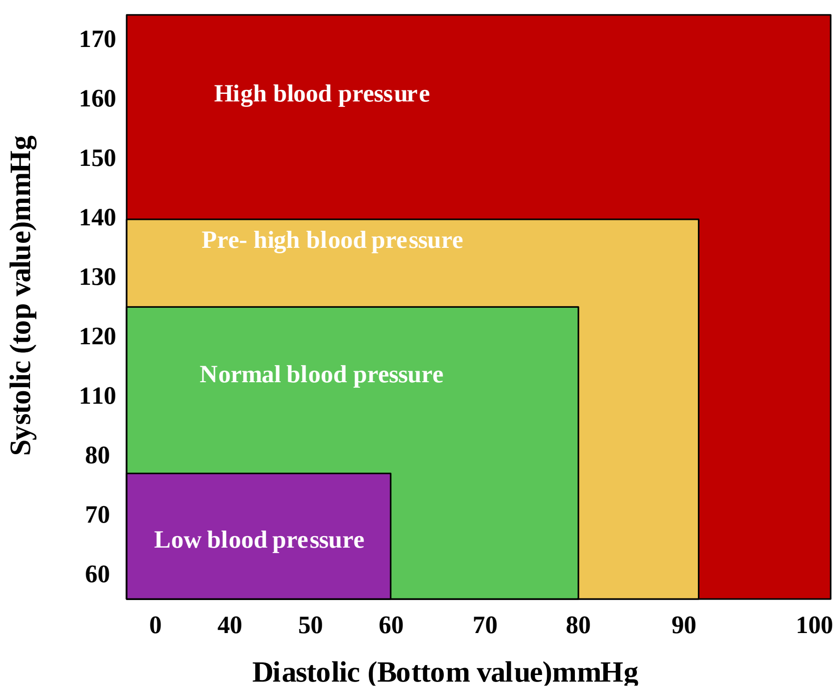


Figure 0-5: La plage de pression artérielle

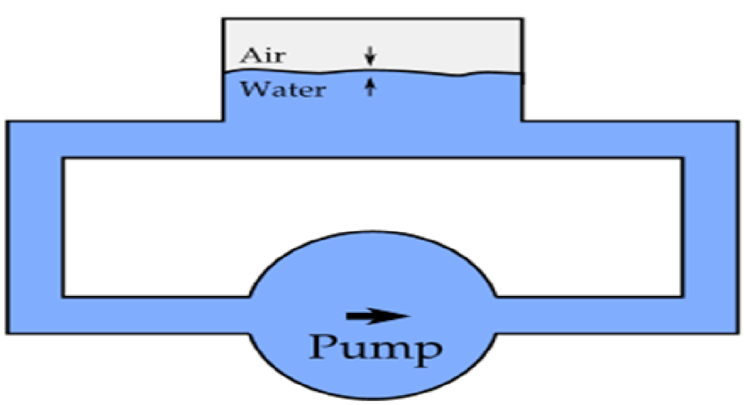
0.6 Représentation cardiovasculaire

La modélisation et les simulations représentent un outil technique puissant dans la pratique. Elle est largement utilisée en raison de son efficacité et de sa variabilité. Appliquée à modéliser le système biologique, elle permet d’étudier un corps dans des conditions physiologiques ou d’évaluer l’effet de modifications pathologiques sur l’organisme et l’impact possible d’un traitement non-invasive.

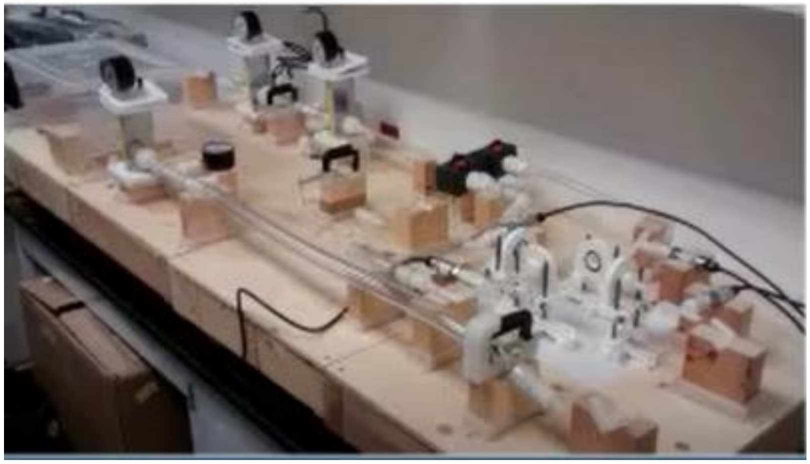
0.6.1 Modelé Mécanique

Certaines études décrivent le cœur comme une pompe mécanique et expliquent le rythme cardiaque par une expression mécanique. Le modèle le plus populaire était le modèle Windkessel, suivi de plusieurs recherches. Le physiologiste allemand Otto Frank a décrit le modèle de Windkessel (1899) [9]. Depuis lors, il est utilisé dans certains modèles physiologiques pour simuler l'élasticité d'un compartiment. Le terme Windkessel se traduit vaguement en allemand par "chambre à air". Le modèle original repose sur un circuit hydraulique constitué d'une pompe à eau raccordée à une chambre d'air, comme illustré à la (Figure 0-6) (a). Les pompes à eau dans la chambre l’eau comprime l’air qui pousse l’eau vers la pompe. Ce processus établit l'équivalent du mouvement du sang sur le corps.

Sur la (Figure 0-6) (b), un modèle mécanique récent, utilisé actuellement pour vérifier l'exactitude de la description mécanique par rapport au système réel, semble fournir une approximation acceptable, mais manque de flexibilité pour changer les tubes si les paramètres étaient modifiés



(a)



(b)

Figure 0-6: (a) Un circuit hydraulique qui consiste en une pompe à eau connectée à une chambre d'air, (b) Un modèle mécanique du système circulatoire

0.6.2 Description Électrique

Bien que l'approche mécanique présente une bonne estimation, elle définit mal certaines caractéristiques délicates de la circulation sanguine dans les vaisseaux. Les études vont plus loin et établissent un modèle de circuit électrique pour obtenir une description des propriétés nécessaires et de la dynamique du système avec un minimum de complexité.

La (Figure 0-7) illustre l'analogie entre le flux sanguin dans les artères systémiques et le courant dans un ensemble d'équations mathématiques électriques entre certaines des variables du système. Le tableau suivant montre les variables les plus courantes entre les deux systèmes et leurs équivalences [10].

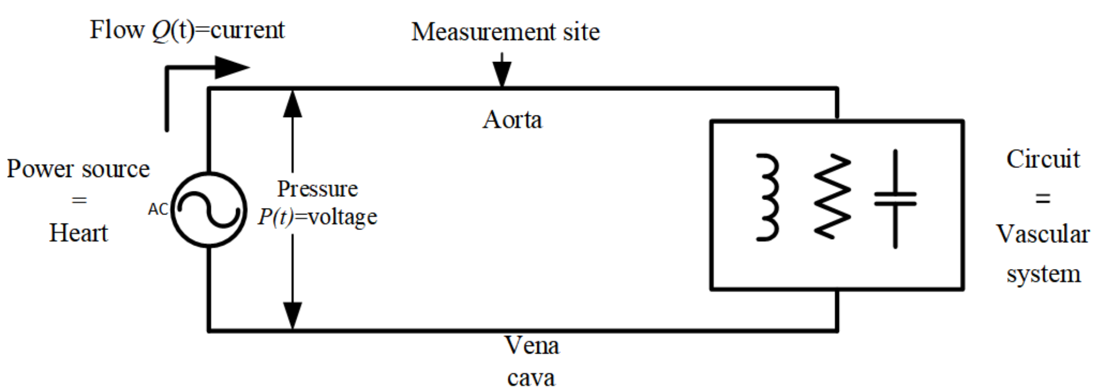


Figure 0-7: Une analogie électrique entre le courant électrique et le flux systémique.

Table 0-1: Analogie entre le model dynamique et électrique

Dynamique des fluides	Physiologie	Analogie électrique
Pression P [pa=J/m3]	Pression artérielle [mmHg]	Tension U [V=J/C]
Débit Q[m3/s]	Débit sanguin [L/s]	Current I [A=C/s]
Volume V [m3]	Volume sanguin [L]	Charge q[c]
Viscosité µ	Résistance sanguine. $R = \frac{8\,\mu N l}{\pi r^4}$	Résistance Electrique R
Coefficient élasticité	Vessel's wall compliance	Capacitance C
Inertie	Inertie du sang	Inductance L
Poiseuille's	$Q = \frac{\Delta P}{R} = \frac{\Delta P \pi r^4}{8\mu l}$	Loi d'Ohm: I=$\frac{\Delta U}{R}$

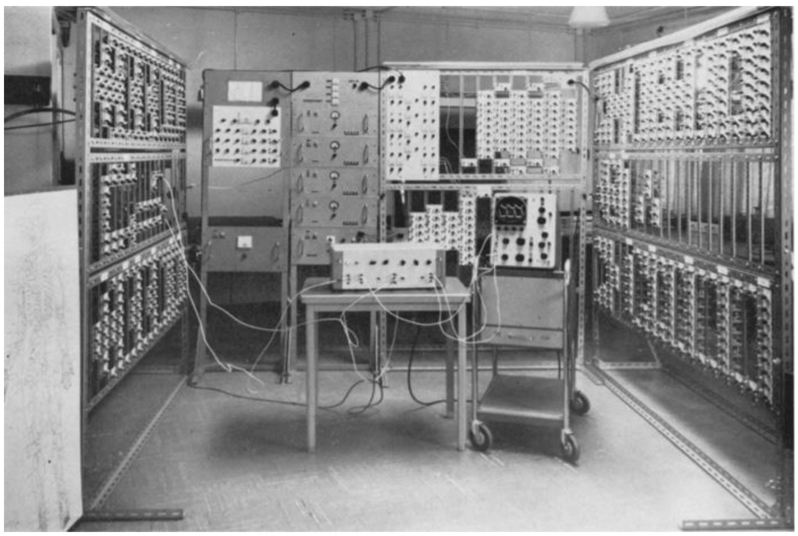


Figure 0-8: Illustration électrique du système cardiovasculaire du corps humain (1966)

Dans la (Figure 0-8), l'analogie proposée par Pater [11], encore utilisait dans les recherches récentes, fournissait une représentation acceptable du système cardiovasculaire. Le chapitre suivant examinera certains modèles qui utilisent la représentation électrique et comment ils estimaient les paramètres

0.7 Modèles d'éléments localisés

Les modèles basés sur les éléments localisés sont utilisés pour estimer les propriétés artérielles totales en ajustant le modèle à la pression et au débit mesurés (aortiques). Un modèle mathématique d'un système physique, qui est simplifié à partir d'éléments spatialement distribuées dans une variable à une seule échelle, est appelé modèle à paramètres/éléments localisés.

Le vrai apport pour le cœur est le débit sanguin du ventricule gauche, qui est délivré par coup en fonction du temps, avec un battement fréquent de 60 sur 70 par minute (environ 0,8 seconde). Sur la base de la courbe de signal de mesure expérimentale montrée sur [25], illustrée à la Figure 0-9a), la courbe de données réelles a été simulée dans certaines conditions (f = 1,29 Hz) avec ADS, comme illustré à la (Figure 0-9b).

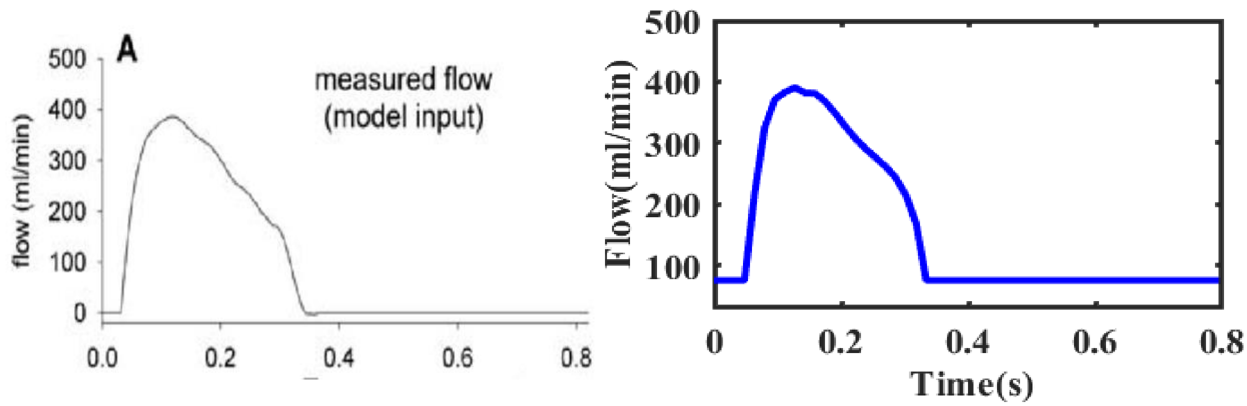
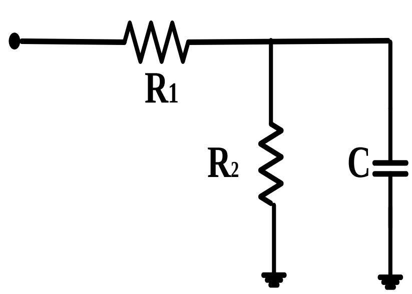
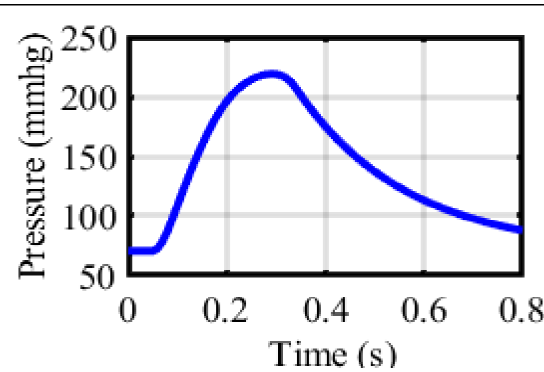


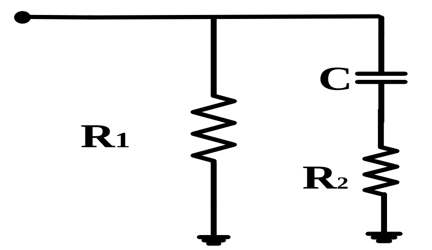
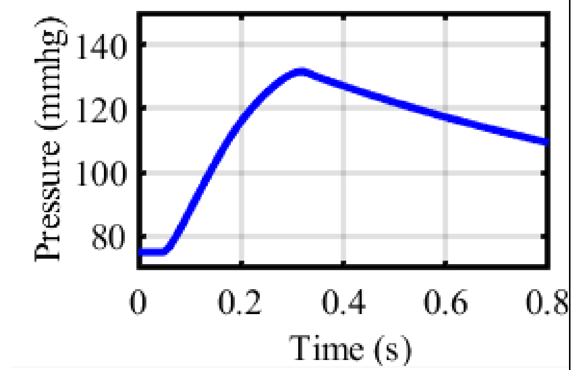
Figure 0-9: a) Débit mesuré b) débit simulé sur ADS

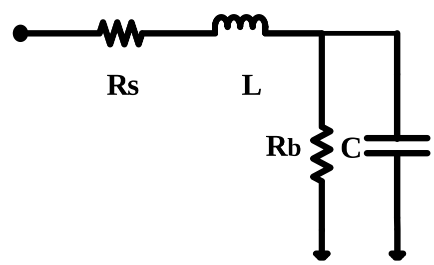
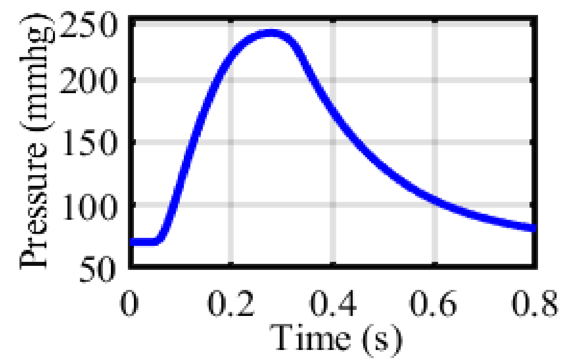
Différentes configurations de modèle coexistent. Dans le tableau 0-2, les deux premières configurations sont des modèles à trois éléments référencés [25]-[28]. Elles offrent une forme de courbe proche de la pression aortique mais la pression résultante est supérieure à la pression normale. Les configurations 3 et 4 [28] sont plus complexes pour améliorer les résultats de mise en forme de ces modèles et pour prendre en compte davantage les caractéristiques des autres vaisseaux sanguins. La limitation de ces modèles réside dans la dépendance à l'ajustement.

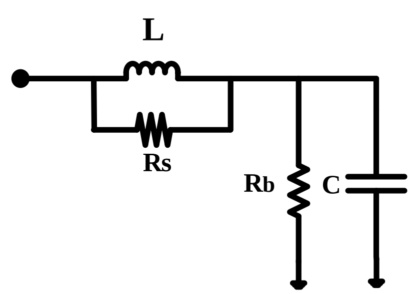
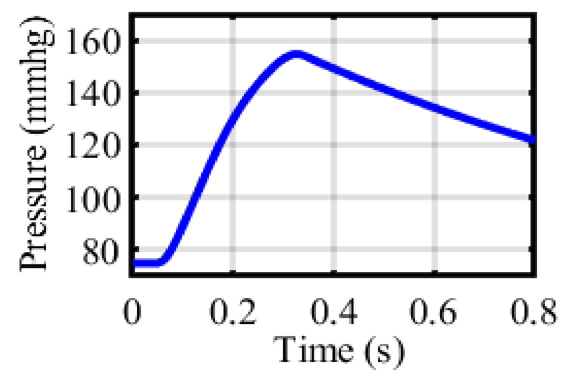
Table 0-2: Certains modèles d'éléments localisés simulés par le logiciel ADS (Advanced Design System) et la pression résultante.

Model	Représentation	Résultat
Trois éléments Windkessel Configuration 1		

<table>
<tr><td>Trois éléments

Configuration 2</td><td>
</td><td>
</td></tr>
<tr><td>Quatre
éléments

Configuration 3</td><td>
</td><td>
</td></tr>
<tr><td>Quatre
éléments

Configuration 4</td><td>
</td><td>
</td></tr>
</table>

D'autres modèles offrent plus de précision en fonction de certaines équations pour évaluer la valeur de chaque élément. Ces équations dépendent des caractéristiques vasculaires et de leur équivalence avec les dépressions de Poiseuille.

La viscosité (µ) : définie comme la résistance des fluides à l'écoulement, la résistance à la circulation sanguine, en particulier pour le sang, inclut les traits sanguins et la forme des vaisseaux [29].

Le flux de fluide : l'énergie de la circulation provient de la différence de pression [30].

La Densité : est défini comme la masse par unité de volume. Le sang est un liquide mélangé et sa masse contient 55% de plasma et 45% de cellules (rouge, blanc et plaquettes), la densité moyenne total du sang chez l'homme étant d'environ (1050-1060) kg / m3 [30].

L'élasticité : Les propriétés des vaisseaux ainsi que l'épaisseur de la paroi des vaisseaux sanguins affecte le comportement et les caractéristiques dynamiques du système circulatoire.

Maintenant le modèle dans [31] utilise 3 éléments C, L, R pour représenter chaque vaisseau, appliquez donc ici les paramètres de l'aorte pour les calculer.

$$\mathrm{R} = \frac{8.\pi.l.\mu}{\mathrm{A}^2} \qquad (1)$$

Où µ est la viscosité du sang, l et A correspondent à la longueur et à la section de chaque segment d'artère.

$$\mathrm{L} = \frac{9.l.\rho}{4.\mathrm{A}} \qquad (2)$$

Où ρ est la densité sanguine

$$\mathrm{C} = \frac{3.l.\pi.\mathrm{r}^3}{2.\mathrm{E.h}} \qquad (3)$$

Où r, E, h sont le rayon de l'artère, le module d'élasticité et l'épaisseur des artères. Les résultats de la simulation effectuée pour une fréquence cardiaque de 1.219Hz sont illustré dans la(Figure 0-10).

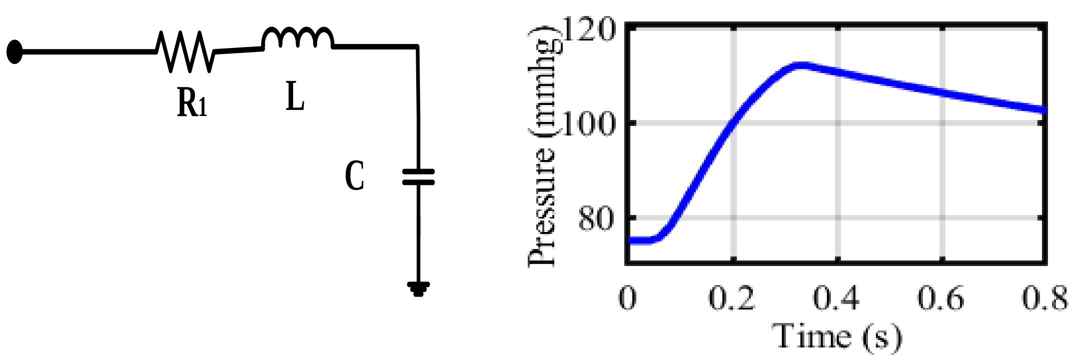


Figure 0-10: Modèle à trois éléments

L'approche décrite dans [33]utilise d'autres équations de Modèle de filtre en π

$$R_S = \frac{8.\mu.l.}{\pi r^4} \tag{4}$$

$$L = \frac{1.33.\rho.l}{\pi r^4} \tag{5}$$

$$C = \frac{l^2}{2\ cp^2 L} \tag{6}$$

$$R_P = \frac{2 * 10^{-6}}{\mathrm{C}} \tag{7}$$

Où c_p est la vitesse de l'onde de pouls, r est le rayon du vaisseau.

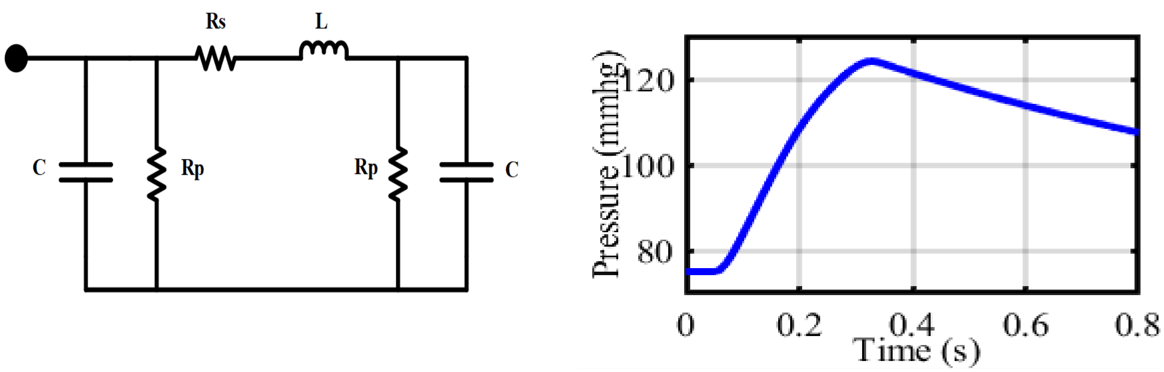


Figure 0-11: Modèle de filtre en π

Le résultat du dernier modèle est plus précis et plus proche de la mesure réelle, mais l'effet du condensateur supplémentaire et de la résistance peut être négligé, et il n'y a pas de grande différence si vous ne faites que simuler avec la charge, le condensateur et la résistance.

Dans la thèse de Pater [11], il a commencé à démontrer l'analogie entre le système hémodynamique et la théorie de la ligne de transmission et a déduit une équation liée à la tension et au courant.

$$R_S = \frac{8.\mu.}{\pi r^4} * l \tag{8}$$

$$L = \frac{\rho}{\pi r^2} * l \tag{9}$$

$$C = \frac{3\pi r^3}{2Eh} * \ l/2 \tag{10}$$

$$R_P = \frac{2 * \mu\, h}{3\pi r^3} * 2/l \tag{11}$$

La précision de ces équations dépend du rapport longueur du segment à la longueur d'onde qui est doit être inferieur a λ / 8.

Tout d’abord, pour démarrer notre circuit initial, la source sinusoïdale à 1 Hz a été utilisée. Auparavant, dans [35] aussi le signal était modélisé comme une onde sinusoïdale d'amplitude I_0 pendant la systole et est égale à zéro sinon.

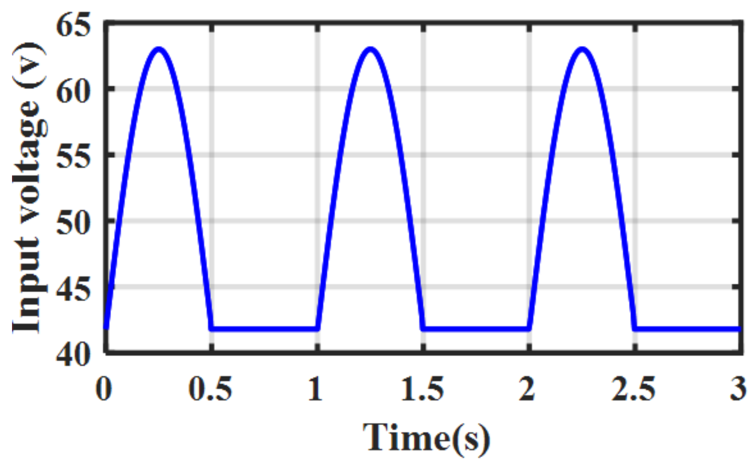


Figure 0-12:Signal d'entrée

Dans le circuit initial, la valeur des éléments a été calculée à l'aide des équations précédentes avec les dimensions d'un segment aortiques, comme indiqué dans [33]. La (Figure 0-13) représente la sortie d’un modèle d’éléments localisés à 1Hz. Pour examiner si ce circuit peut fonctionner à des fréquences plus élevées, l’étude inclura une augmentation de la fréquence avec le nombre de cellules et un réglage des paramètres jusqu’à ce que la sortie soit dans la plage normale de la pression.

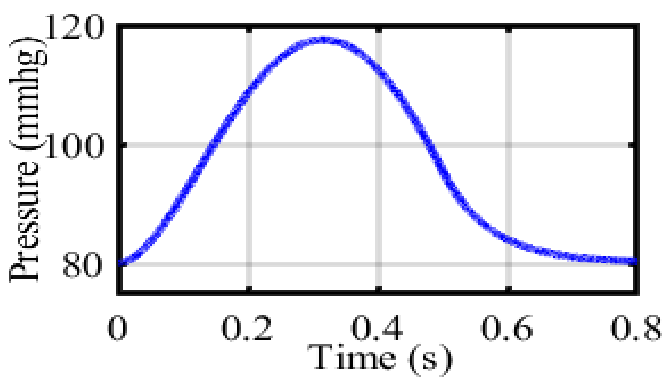
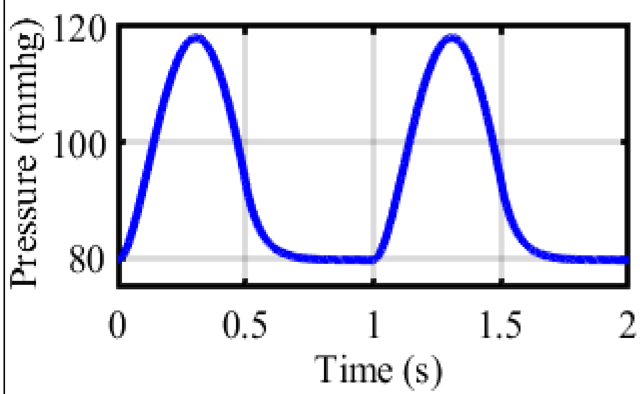
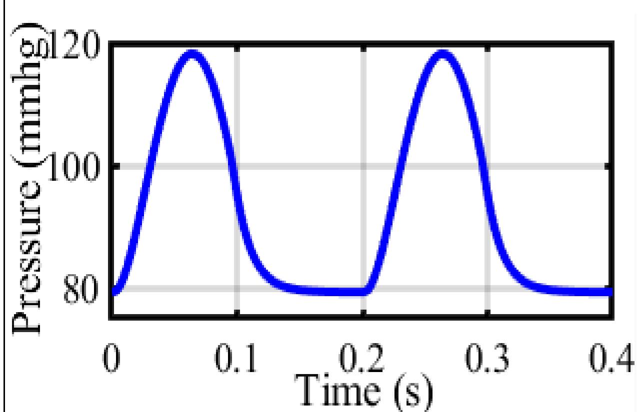
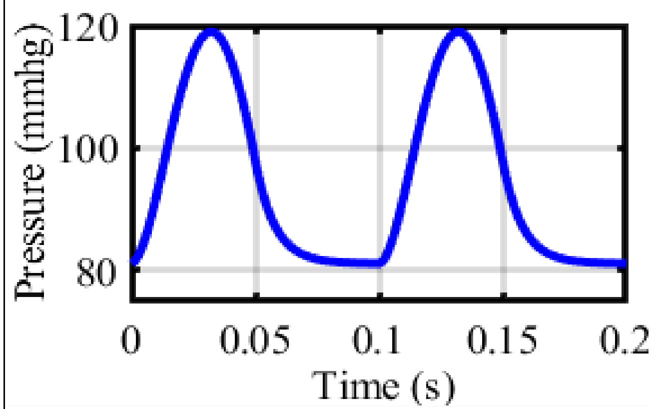
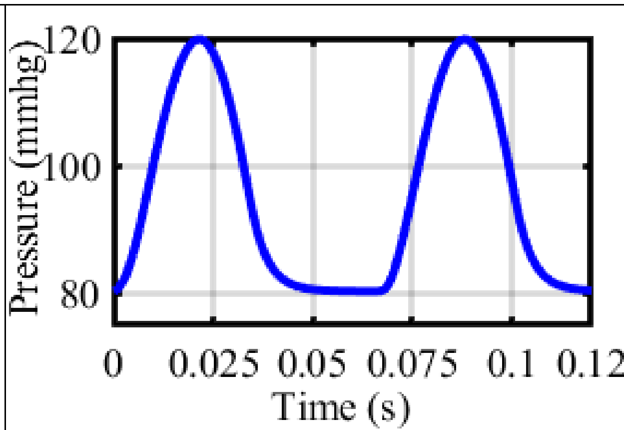
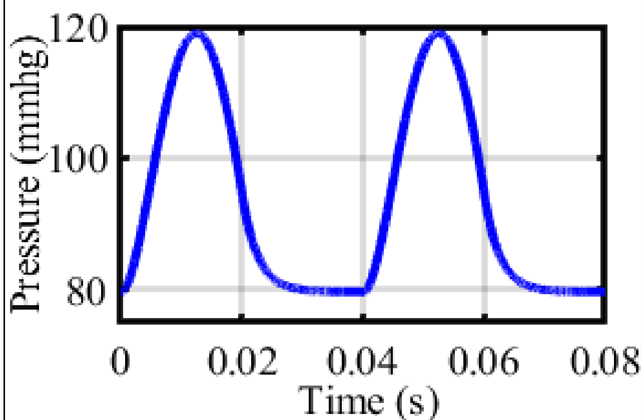
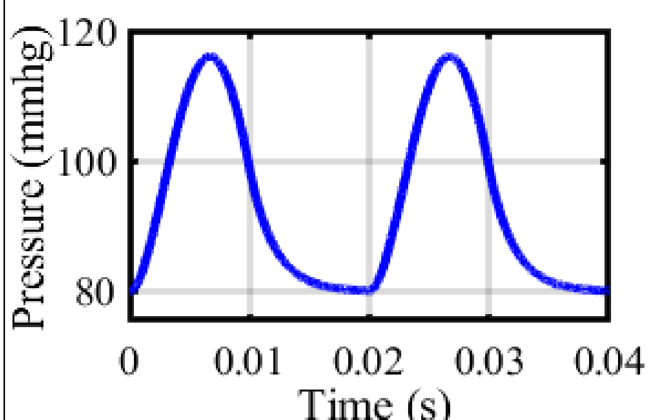
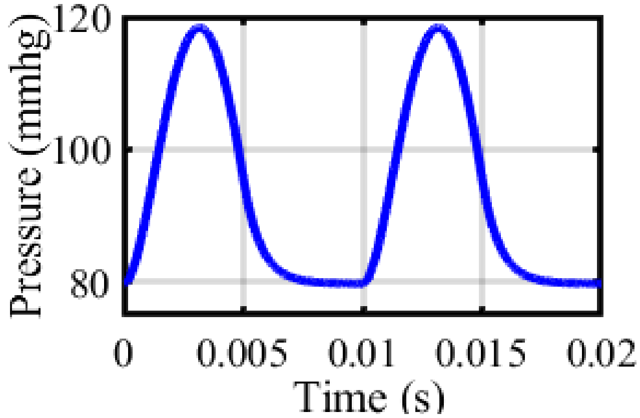


Figure 0-13: Signal de sortie d’un model en π a Hz

Table 0-3: Signal de sortie à chaque fréquence et la valeur de chaque élément

Frequence (Hz)	Pression de sortie (volt –mmHg)	L (H)	C (pF)	R1 (K-Ω)	R2 (K-Ω)
1		3.07e4	6.8	16.7	300
5		1.3e3	6.8	0.5	540
10		307	6.8	0.158	690

15		137	6.8	0.075	818.4
25		52	6.8	0.021	966.2
50		13.87	6.8	0.007	1162.1

100		2.68	6.8	0.0017	1382.2

Comme indiqué dans le tableau précédent, chaque fois que la fréquence est augmentée, les paramètres L, R1, R2 sont influencées par le changement de fréquence. Comme noté, les inductances L et R1 ont diminuées, le C reste constant, mais R2 augmente différemment avec la fréquence.

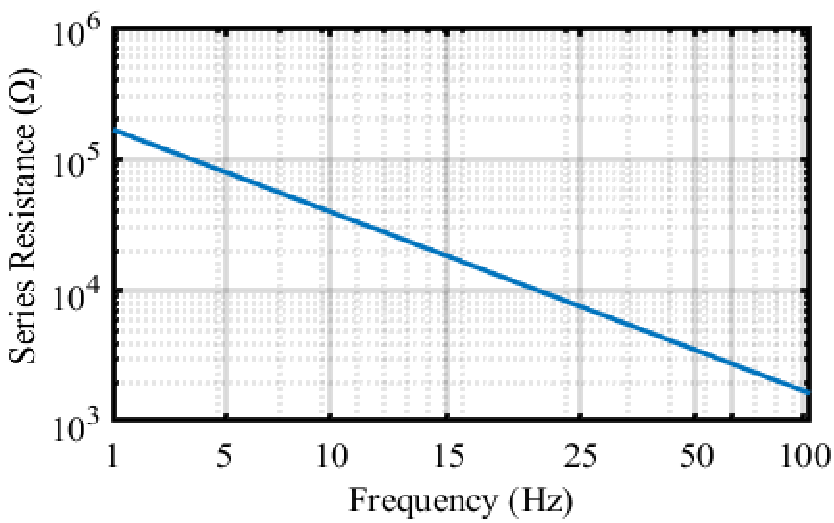


Figure 0-14: Valeur de la Résistance série avec fréquence

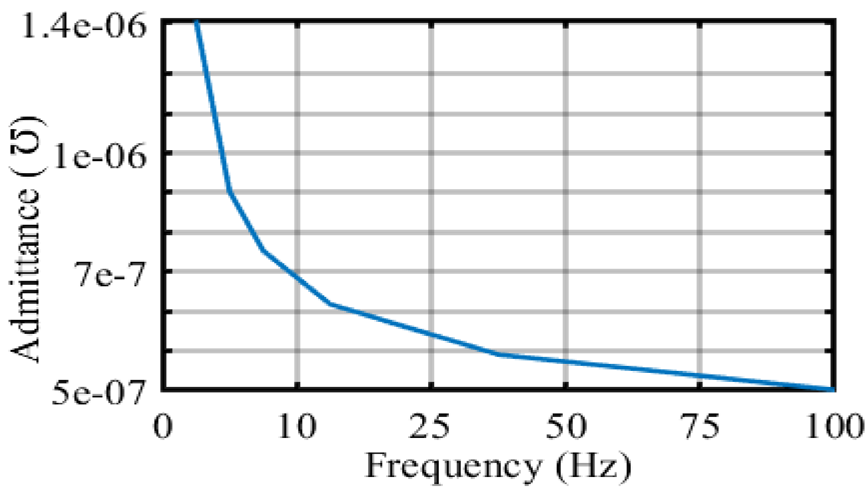


Figure 0-15: Valeur de l'admittance parallèle avec fréquence

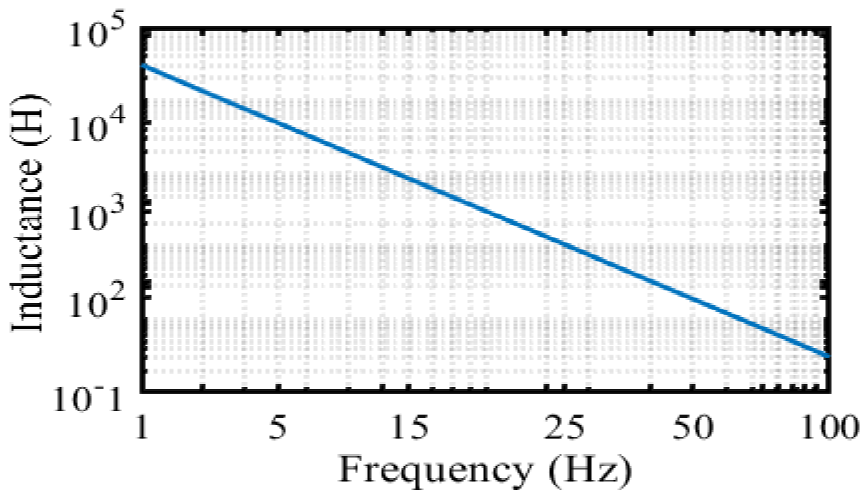


Figure 0-16: Valeur de l'inductance avec fréquence

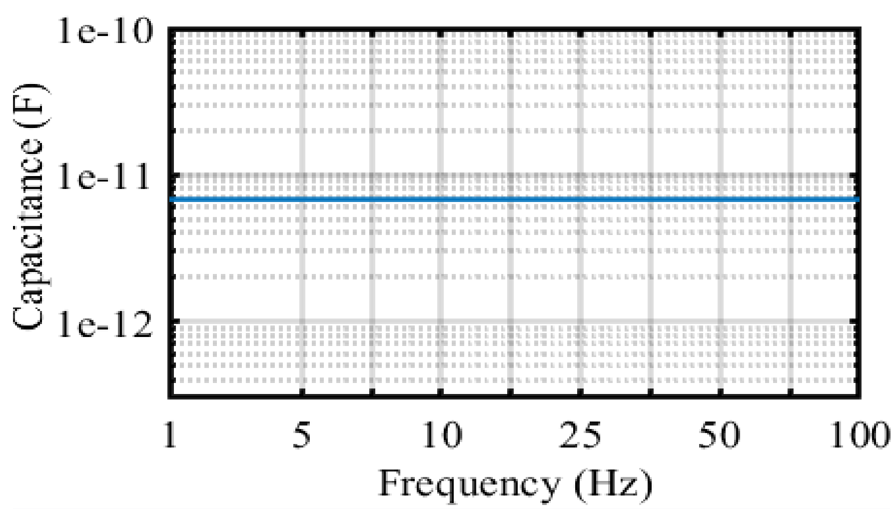


Figure 0-17: Valeur du condensateur avec fréquence

Afin de continuer avec des fréquences plus élevées, le nombre de cellules gênera le processus. Ainsi, les expressions suivantes décrivent la dépendance de chaque élément au changement de fréquence, comme indiqué dans les figures précédentes, à l'aide des formules suivantes :

$$R(f) = \frac{\mathrm{R}_1}{f^2} \tag{12}$$

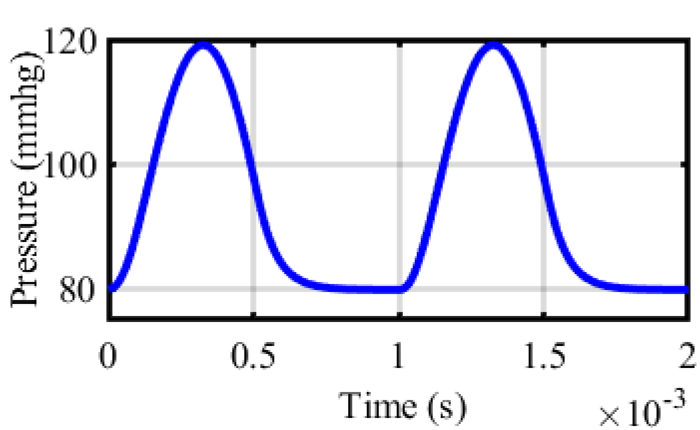
$$Y(f) = Y * \ln(f) \tag{13}$$

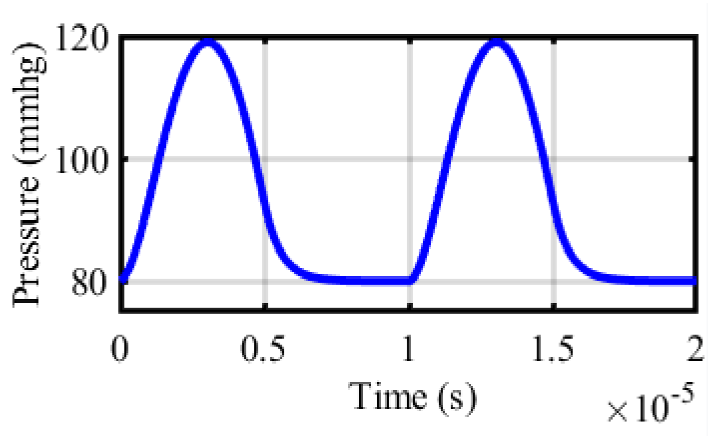
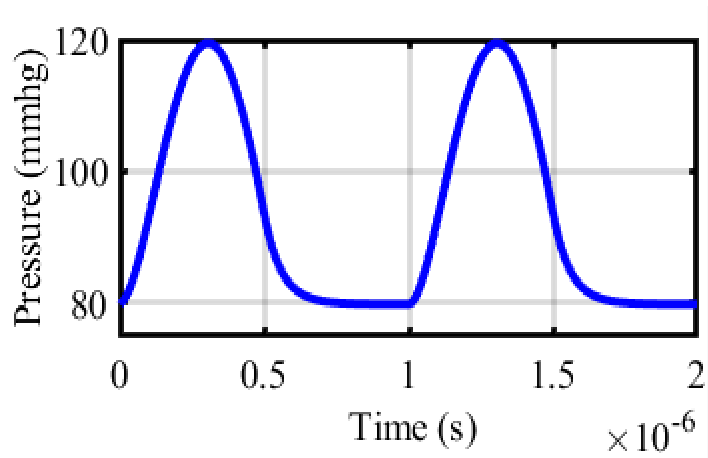
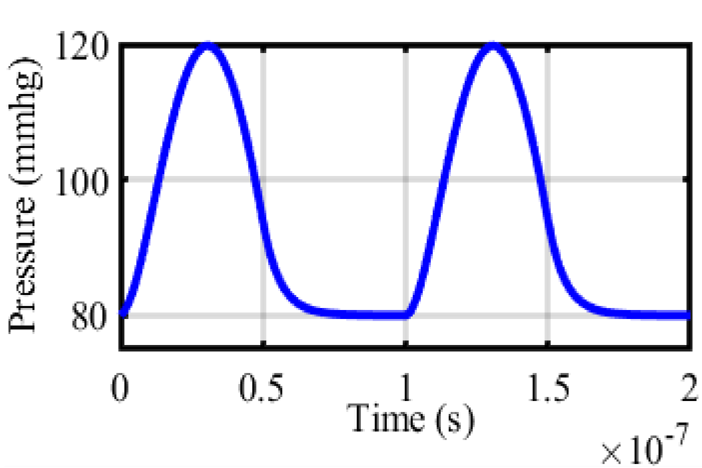
$$L(f) = \frac{L_1}{f^2} \tag{14}$$

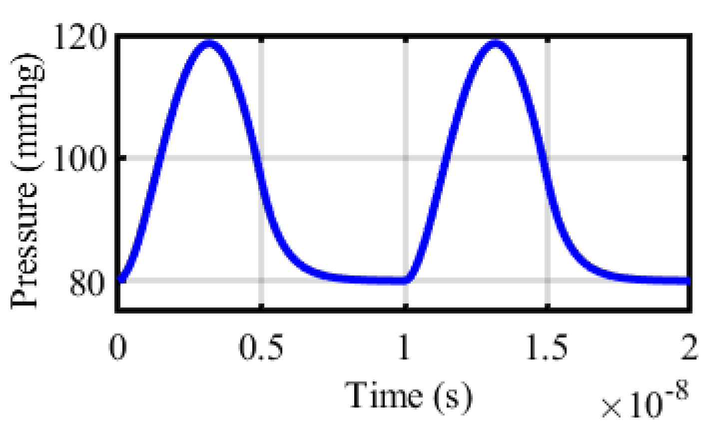
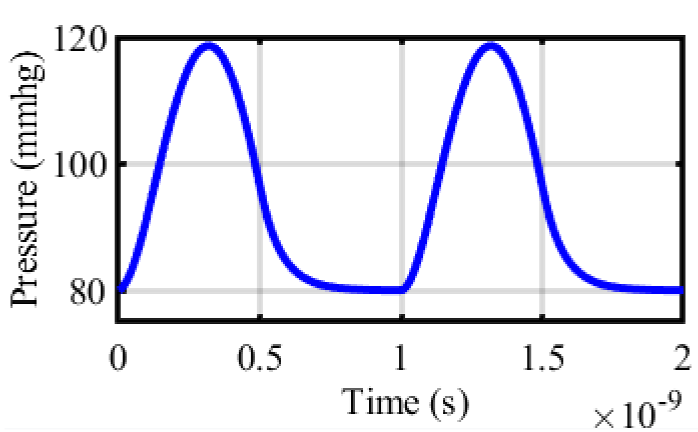
$$C(f) = C1 \tag{15}$$

Ces équations démontrent la relation entre chaque bloc et la fréquence f. En outre, l'influence de ces paramètres sur la forme d'onde pendant le réglage a été remarquée. Maintenant, des fréquences plus élevées peuvent être facilement simulées en utilisant une cellule. Le tableau suivant montrera la sortie à chaque fréquence et validera nos équations en même temps, en considérant (I_{max}) à chaque fréquence.

Table 0-4: La pression de sortie à haute fréquence

Frequency	output
1KHz	

100KHz	
1MHz	
10MHz	

100MHz	
1GHz	

(Figure 0-18) montre les résultats avec l'axe du temps dénormalisé. La correction apportée avec la fréquence conserve le même résultat et démontre la scalabilité avec la fréquence.

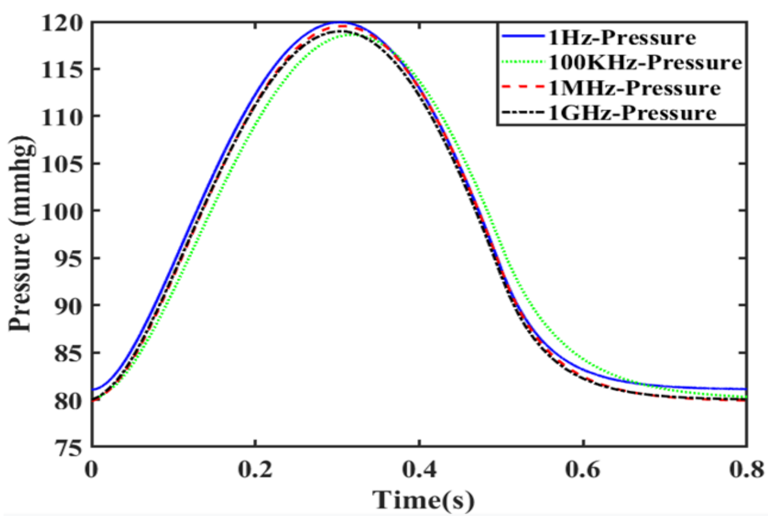


Figure 0-18: Forme d'onde à différentes fréquences dénormalisées

0.8 Modèle Microruban

L'effet de l'augmentation de la fréquence sur les éléments localisés été étudié. Cette étude démontrer le fait que le système circulatoire peut être présenté comme un modèle miniaturisé à haute fréquence, en particulier aux fréquences RF, en tant que ligne de transmission. En raison de la faible longueur effective aux hautes fréquences.

La ligne Microruban est considérée comme la ligne de transmission la plus utilisable en raison de la simplicité de sa miniaturisation et de sa fabrication par des procédés photolithographiques, de son faible coût et de la possibilité de travailler avec différents types d'appareils. Une description pratique simple du microruban est qu'un conducteur de largeur «W» est imprimé sur un substrat diélectrique fin et moulu d'épaisseur «h» et de permittivité relative ε_r [36].

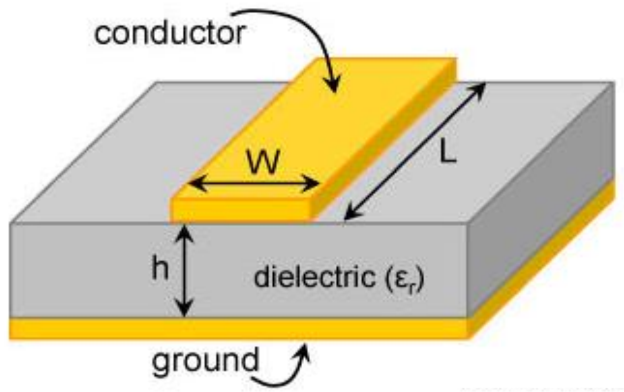


Figure 0-19: Illustration de la ligne microruban

Une ligne de transmission à microruban est illustrée à la (Figure 0-19). Ses caractéristiques physiques comprennent la largeur de la microruban (w), son épaisseur (h), la hauteur du substrat (d) et la constante de permittivité relative (ε_r).

L'observation est que la conception optimale des microrubans devrait être basée directement sur les dimensions physiques. Cela nécessite des modèles et une conception précise ; le modèle doit être sous une forme facilement calculable. Dans la conception proposée, une étude Matlab a été utilisée à différentes εr allant de (2 à 20) et différentes épaisseurs (0,025 à 8 mm) pour définir le meilleur accord aux équations du modèle entre la permittivité relative, l'inductance et la capacité. Hammerstad a fourni des formules bien acceptées pour calculer la permittivité effective et l'impédance caractéristique des lignes à microruban [38][39]:

$$A = \frac{Z0}{60}\sqrt{\varepsilon_r + 1/_2} + \frac{(\varepsilon_r - 1)}{\varepsilon_r + 1} * (0.23 + \frac{0.11}{\varepsilon_r}) \tag{16}$$

$$B = \frac{377\pi}{2Z0\sqrt{\varepsilon_r}} \tag{17}$$

Le cas du w <2

$$w = \frac{8e^{A}}{e^{2A}} - 2 \tag{18}$$

Dans le cas où w> 2, une autre formule sera utilisée,

$$w = \frac{2}{\pi(B - 1 - \log(2B - 1)} + \frac{\varepsilon_r - 1}{2\varepsilon_r \log(B - 1)} + 0.39 - \frac{0.61}{\varepsilon_r} \tag{19}$$

La permittivité effective ε_{eff} peut être calculée comme suit :

$$\varepsilon_{eff} = \frac{\varepsilon_r + 1}{2} + \frac{\varepsilon_r - 1}{2} * \frac{1}{\sqrt{(1 + 12 * w/h)}} \tag{20}$$

Et les L et C des formules suivantes :

$$L = \frac{Z0}{c0 * \sqrt{\varepsilon_{eff}}} \tag{21}$$

$$C = \frac{1}{c0 * Z0 * \sqrt{\varepsilon_{eff}}} \tag{22}$$

0.9 Etude d'impédance

Dans la plupart des lignes de transmission, les effets dus à L et C tendent à être dominants en raison de la résistance en série et de la conductance de dérivation relativement faibles. Au chapitre trois, la relation entre l'élément forfaitaire et la fréquence est présentée comme suit :

$$L(f) = \frac{L_1}{f^2} \tag{23}$$

$$C(f) = C1 \tag{24}$$

Dans ce cas d'impédance sans perte sera :

$$Z = \sqrt{\frac{L1/_{f^2}}{C1}} \tag{25}$$

En d'autres termes,

$$Z = \sqrt{\frac{L1}{f^2 * C1}} = \sqrt{\frac{L1/_{f}}{f * C1}} \tag{26}$$

A partir de ces relations on peut supposer une équation plus général L (f) = L1 / (f * α1) ou la valeur maximale de α1 est de f et la valeur minimale est de 1. Supposons aussi que *C* est multiplié par α2 égal à 1 lorsque α1 est f, il sera multiplié par f lorsque α1 est 1. Par conséquent, l'impédance est flexible et peut suivre de nombreuse valeur mais sous une condition maximale de α1 *α2 = f. Pour vérifier ce résultat, une étude paramétrique a été réalisée pour valider à la fois Z et β.

Lorsque les définitions précédentes d'impédance caractéristique sont utilisées dans l'étude paramétrique, la condition explicite de la meilleure impédance caractéristique appliquée aux équations de la ligne de transmission sera alors évaluée, ainsi que les valeurs des paramètres prédit [40].

0.10 Estimation du modèle Microruban

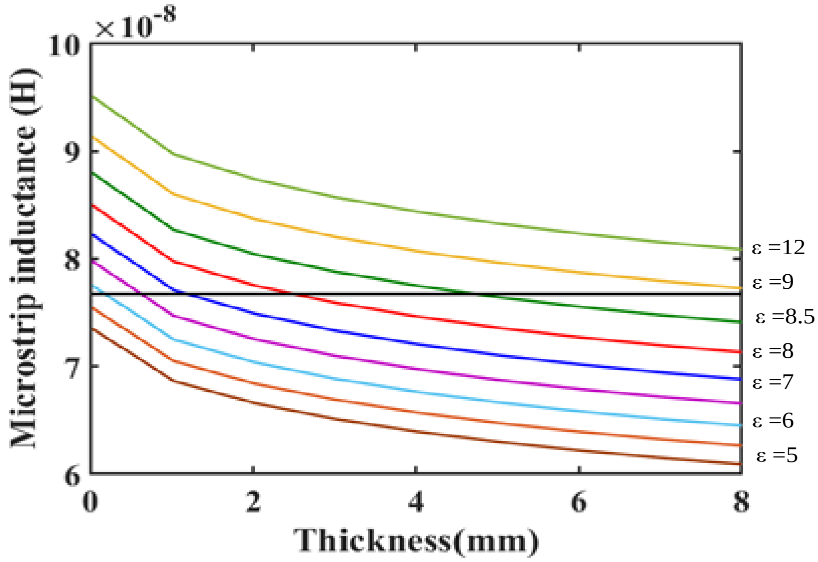


Figure 0-20: Inductance de ligne Microruban à différentes permittivités ε avec l'inductance optimale du modèle

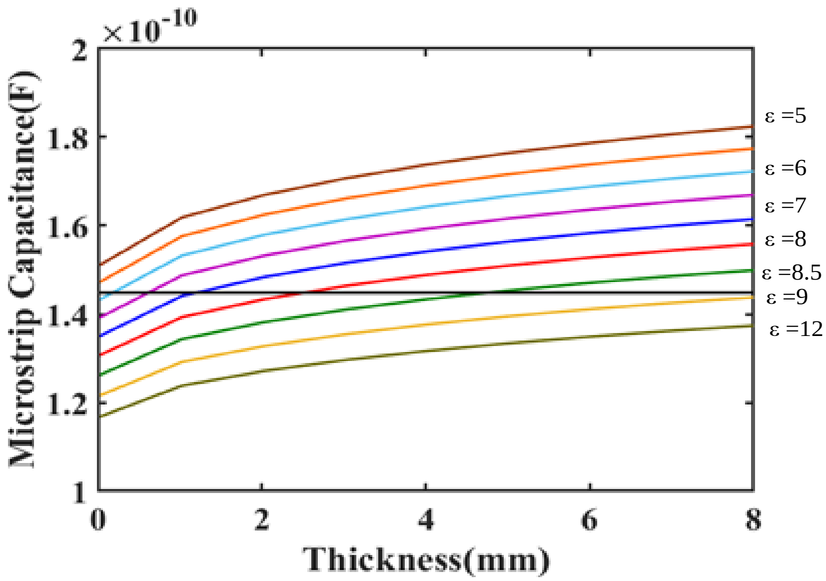


Figure 0-21: Capacité de ligne Microruban à différentes permittivités ε et capacité optimale du modèle

Les(Figure 0-20) représentent l'inductance du Microruban à des permittivité différente et l'inductance du modèle. Cette inductance idéale correspond à plusieurs valeurs de substrat (permittivité/épaisseur). La même manière a capacitance idéale de la (Figure 0-21), c'est-à-dire qu'il existe plus d'une permittivité admise pour la fabrication avec des épaisseur différente.

Pour déterminer la meilleure épaisseur, on a étudié R1 du modèle avec la résistance de surface de l'équation (33). La (Figure 0-22) montre que l'épaisseur de 2.8 mm est l'épaisseur appropriée.

$$R_S = \sqrt{\omega\mu / 2\sigma} \tag{27}$$

La résistivité superficielle des parois de cuivre est (σ = 5,8 × 107 S / m)

Après avoir défini l'épaisseur requise, il convient de calculer la largeur de la ligne de bande.

La formule pour une plaque parallèle sera utilisée, divisée par deux pour avoir la formule du Microruban.

$$G = \frac{\omega\epsilon'' w}{d} \tag{28}$$

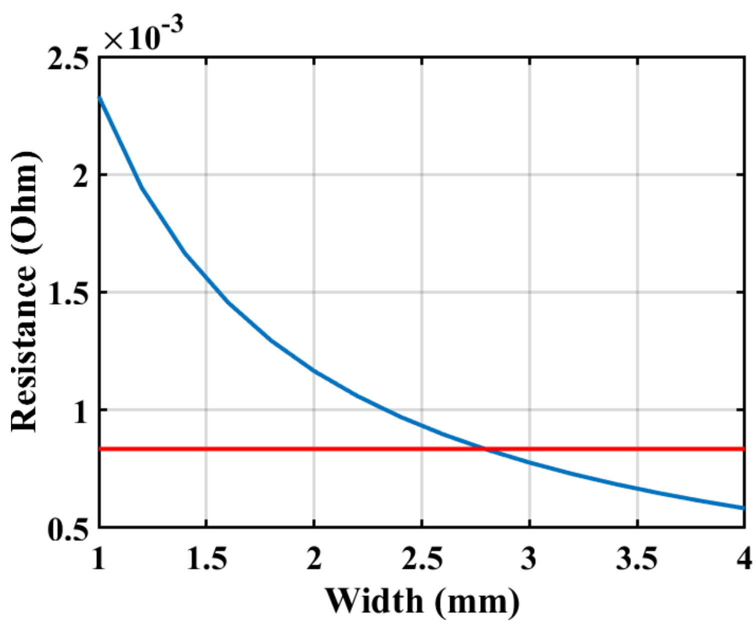


Figure 0-22: Résistance du Microruban à différentes largeurs et résistance du modèle

À partir des équations (21), (22), (27) et (28), le paramètre requis de ce modèle est ε_r = 8, la largeur du substrat égale à 2.8 mm et l'épaisseur de conducteur sélectionnée égale à 0,065 mm avec une tangente de perte = 0.0024.

Le substrat le plus proche est le RT / Rogers 6010. Après application de ces paramètres à la simulation ADS, ils donnent un résultat acceptable, comme indiqué dans la figure suivante.

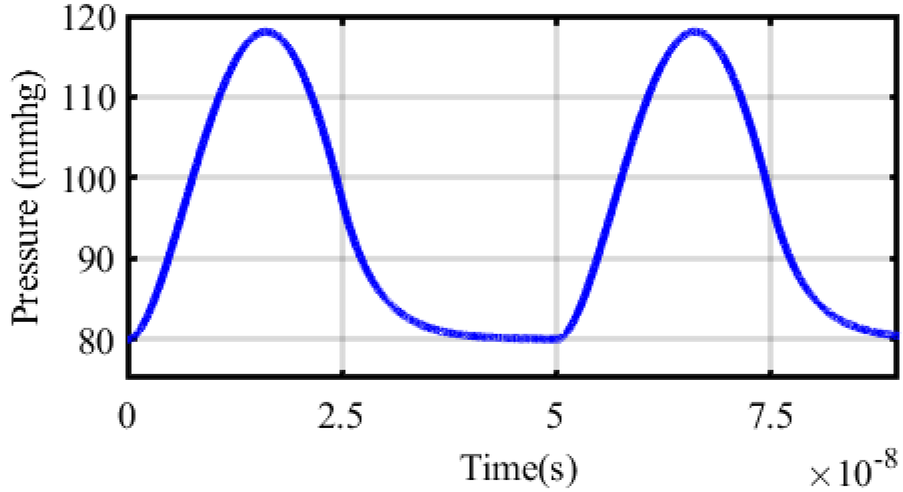


Figure 0-23: Resulat ala sortie du modele microruban

Pour valider la variation de la vitesse d'écoulement dans le vaisseau sanguin, le substrat sera examiné à différents c_p et différents rayons. Les figures (0-26 et 0-27) illustrent les résultats pour confirmer l'aptitude à utiliser le même substrat avec des diamètres différents.

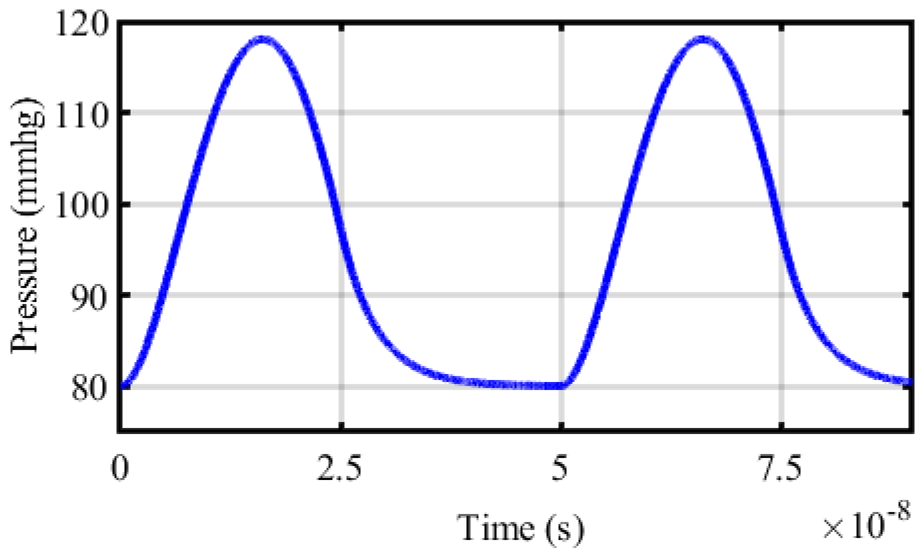


Figure 0-24: Résultat en utilisant le substrat à c_p = 12 et r = 3mm

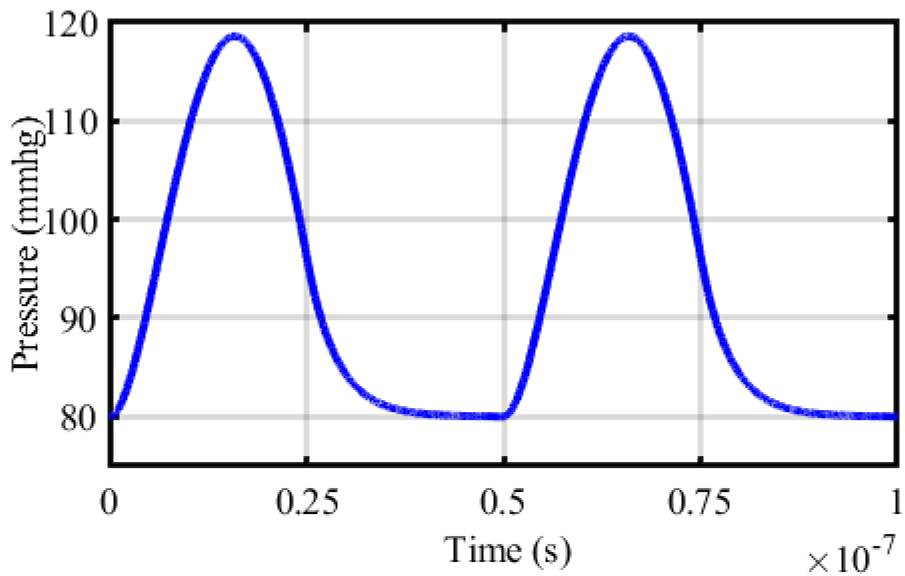


Figure 0-25: Résultat en utilisant le substrat à c_p = 18 et r = 1mm

La fabrication du Microruban proposé a été réalisée avec les équipements de laboratoire.

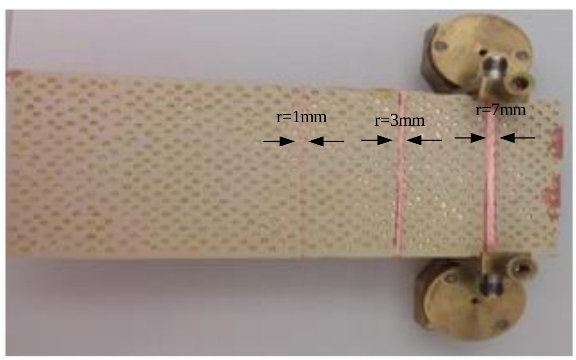


Figure 0-26: Microruban fabriqué représentant différents vaisseaux

Le signal d'entrée a été effectué à l'aide d'un générateur d'impulsions (81133A) à 20 MHz, présenté à la (Figure 0-27) avec le signal d'entrée obtenu utilisé.

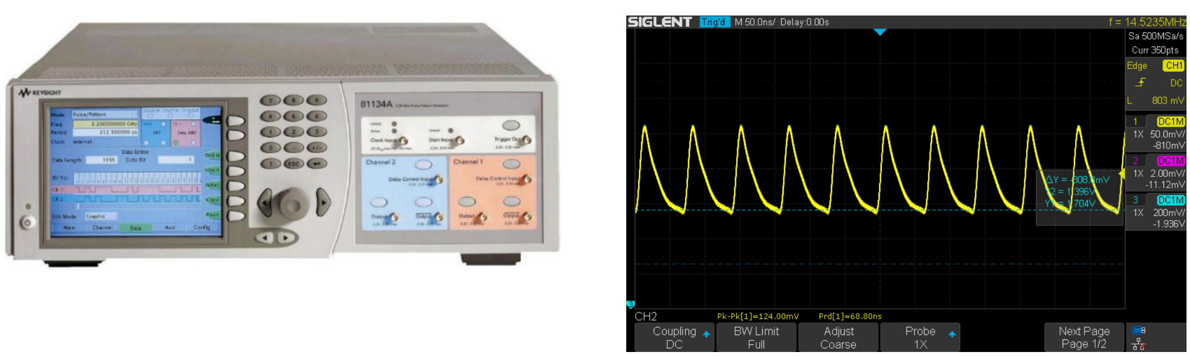


Figure 0-27: Le générateur d'impulsions (81133A) le signal d'entrée présenté sur l'oscilloscope

La(Figure 0-28) montre le résultat correspondant lorsque le rayon est égal à3 mm, même le signal a du bruit mais le signal obtenu est considéré comme un résultat acceptable. La limite vient plutôt du signal d'entrée.

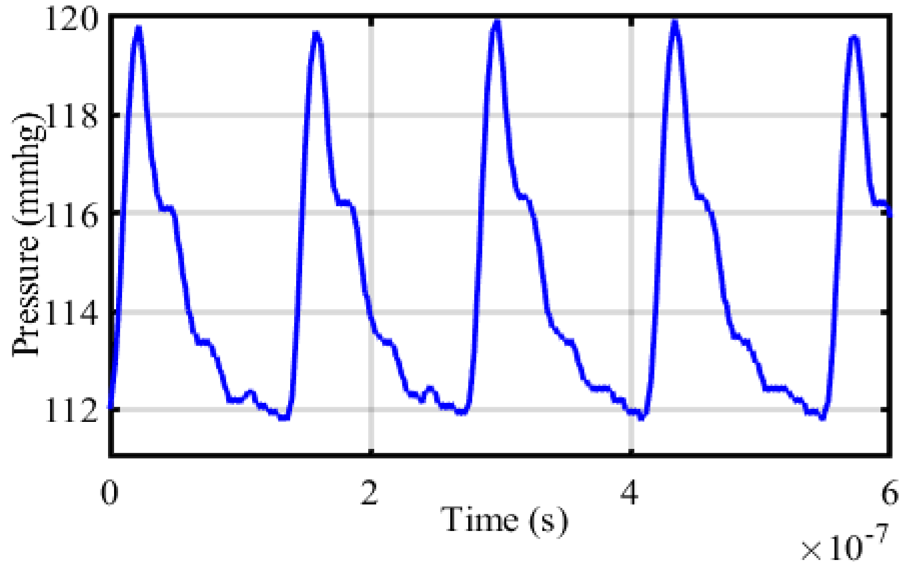


Figure 0-28 : Le résultat obtenu avec un Microruban qui correspond à un vaisseau r = 3mm

En outre, le modèle a été mesuré en tant que données de paramètre S et la (Figure 0-29) montre le résultat obtenu par Advanced Design System (ADS).

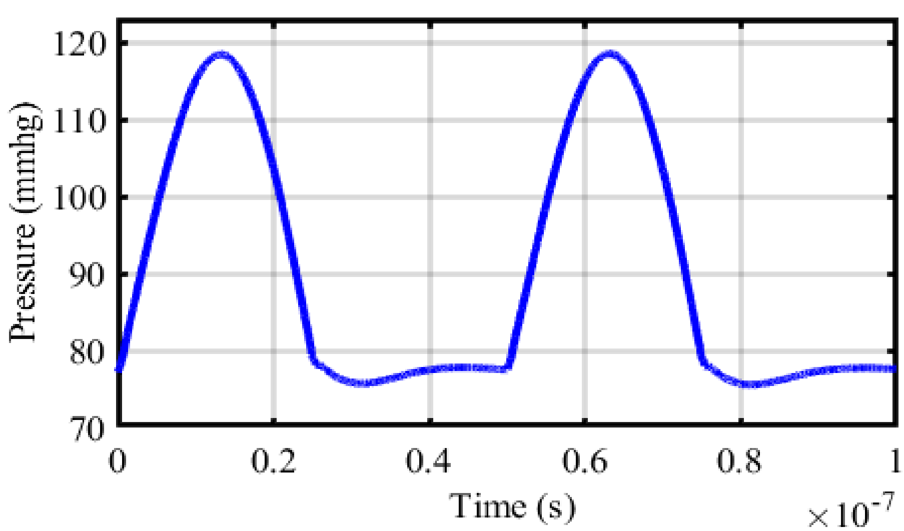


Figure 0-29: Pression simulée à partir des paramètres S mesurés

0.11 Conclusion et perspective

Cette recherche présente un modèle en ligne de transmission RF des vaisseaux du système circulatoire en utilisant le logiciel ADS. Un modèle miniaturisé du système circulatoire humain a été élaboré, basé sur l'équivalence du système circulatoire et de la théorie de la ligne de transmission.

Notre recherche a démontré la scalabilité avec la fréquence en vue de miniaturisé le modèle et par la suite une étude paramétrique a permis la sélection d'un substrat et les dimensions d'une ligne Microruban équivalente. Une fois toutes ces étapes terminées, les simulations du modèle et les mesures initiales indiquent une pression artérielle acceptable dans l'aorte. Ce modèle sera un excellent outil pour comprendre les mécanismes de la circulation sanguine. Il constitue donc la première étape pour montrer le système dans son ensemble.

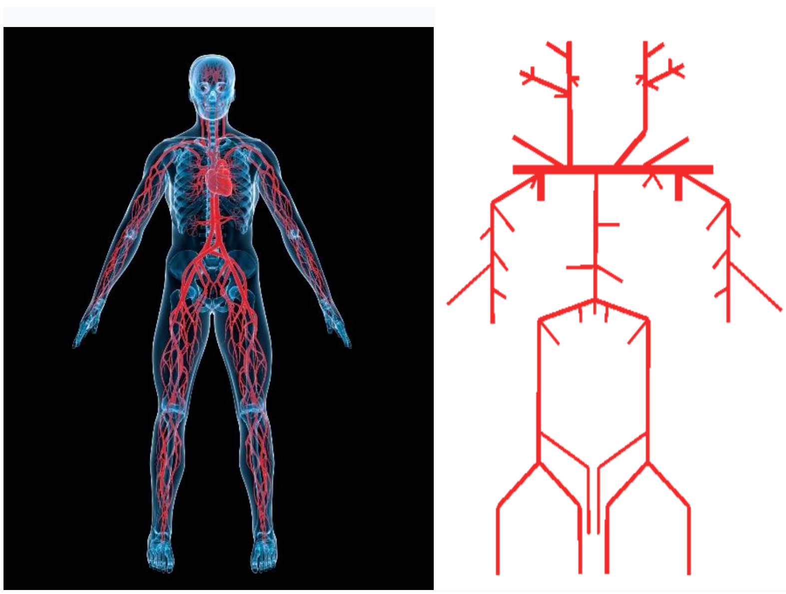


Figure 0-30 système circulatoire complet utilisant la disposition de la ligne de transmission

Les travaux futurs consistent à concevoir l'ensemble du corps humain à l'aide de la ligne de transmission en fonction des données cliniques présentées à la (Figure 0-30). Premièrement, la partie passive comprend l'ensemble du système circulatoire et du système nerveux, et la partie active comprenant le cœur, le foie et les autres organes. En outre, la modification de la forme d'onde de pression peut être un signe de la définition d'un problème de santé. L'effet de chaque élément isolé indique un problème spécifique. Expérimentalement, après avoir utilisé le réglage, nous remarquons que le changement de R1 affecte le cycle de la systole en même temps que le cycle de la diastole influencé par L. À partir de cette information, les maladies peuvent être diagnostiquées après application de la théorie de la ligne de transmission sur le modèle conformément à une conception complexe complète prise en charge par le réseau artificiel.

TABLE DES MATIÈRES

LIST OF FIGURES

LIST OF TABLES

LISTE DES SIGLES ET ABRÉVIATIONS

ADS	Advanced Design System software
A-V node	AtrioVentricular node
S – A node	SinoAtrial node
A-V valve	AtrioVentricular valve
RC- Circuits	Circuit with both a resistor (R) and a capacitor (C)
mmHg	The millimeter of mercury high at 0°C and under the acceleration of gravity
WK-3	The three-element Windkessel
WK-4	Four –element Windkessel
cp	Pulse wave velocity
μ	Blood viscosity
ρ	Blood density
ε_r	Relative permittivity
ε_{eff}	Effective permittivity
σ	Surface resistivity

INTRODUCTION

This chapter will provide background information about some basics of the anatomy of the cardiac and circulatory system.

1.1 Circulation system

The circulation system is one of the most complicated systems in the human body; this system is responsible to deliver the blood, nutrients, oxygen, and other hormones from and to the body cells. Many researchers are trying to explain the structure and functionalities of this system [1] [2].

Mainly it consists of three parts: cardiovascular (the heart), pulmonary (lung), systemic (arteries, veins, and coronary) that jointly work in an organized and systemic manner [2]. The cardiovascular part involves the heart, the blood vessels, and the blood. The heart is the most significant organ in the circulatory system and its importance related to its function because it continuously pumps the blood through the body and generates blood flow. The blood flows in a network of tubes called vessels. Blood is the fluid that transports nutrients, oxygen, carbon dioxide, and many other materials through the body.

1.1.1 The heart

The heart is located slightly to the left of the middle of the chest and the walls of the heart are made of a unique type of muscle called cardiac muscle. Cardiac muscle cells are arranged in a network that allows the heart to contract and relax rhythmically and involuntarily without becoming fatigued [3].

The heart has four chambers as shown in (Figure 1-1), two on the left side and two on the right side. The top chambers called atrium, and the bottom chambers called ventricles. The atria fill with the blood returning from the body or the lungs. The ventricles are responsible to pump out the blood to the body after receiving the blood from the atria. The atria and ventricle are separated by the atrioventricular (A-V) valve, (Bicuspid valve) on the left side and (Tricuspid valve) on the right side [3] [4].

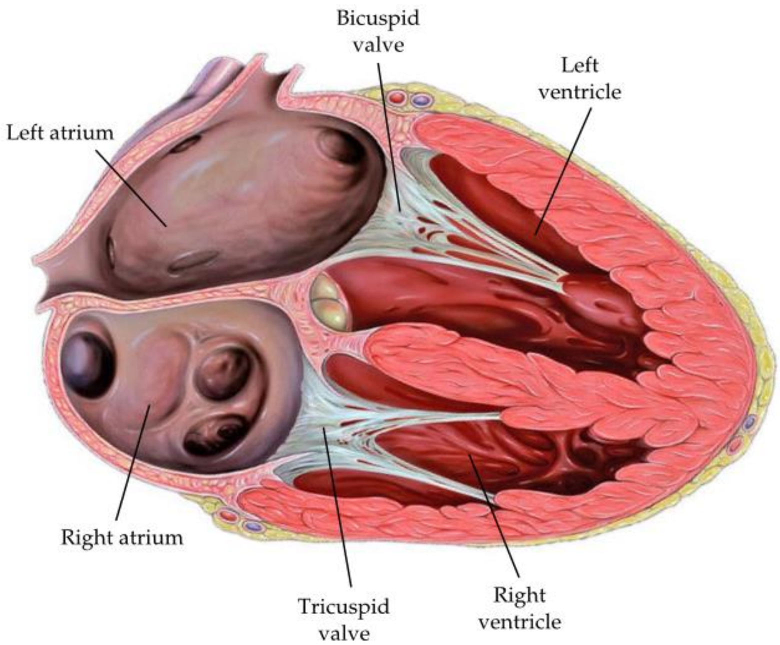


Figure 1-1: An illustration of a sectioned heart [6].

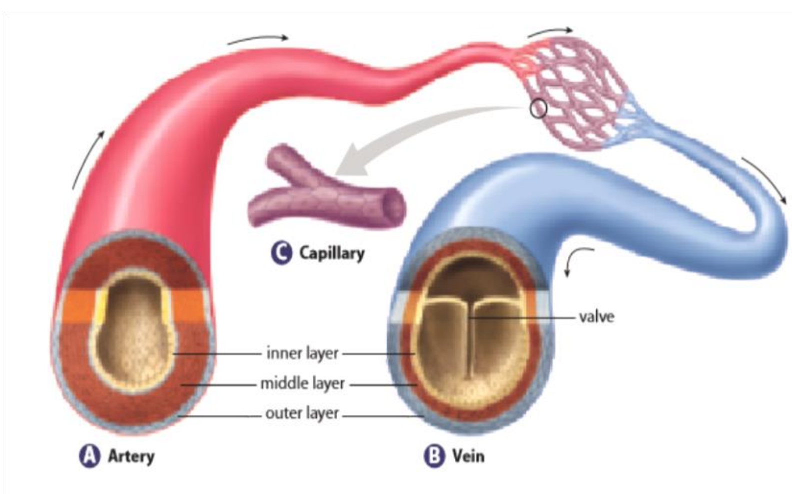


Figure 1-2: Arteries (A) and veins (B) and representing of the three layers. (C) Capillaries consist of a single layer of cells.

1.1.2 Blood vessel:

There are three types of vessels illustrated in (Figure 1-2) and each vessel has its function depending on its position and properties.

1.1.2.1 Arteries:

Arteries are carrying blood from the heart to the body. This kind of vessels has an elastic wall, which expands subject to the blood flow from the ventricles during the contraction, and compresses when the ventricles are relaxing. The aorta (the left side), and pulmonary artery (the right side) are the main arteries that leave the heart [2] [3]. The aorta is the responsible to proceeds the oxygenated blood from the left side of the heart to the rest of the body, and the pulmonary artery has the reverse operation which supplies the deoxygenated (poor- oxygenated) blood to the lungs from the right side of the heart to be re-oxygenated [5].

1.1.2.2 Veins

Veins are the blood transporters toward the heart. Their walls thinner than arteries and are not as elastic as the arteries, which is not helpful for the movement of the blood. Instead of that, veins have one-way valves that prevent the blood from flowing backward. These valves importantly act in the lower body because they ensure that the blood flows upward to your heart against the downward pull of gravity [3]. Vena Cava is the main vein [4], smaller-diameter arteries are called arterioles, and smaller-diameter veins are called venues.

1.1.2.3 Capillaries

Capillaries are the smallest blood vessels and have a very thin wall to permit the oxygen and nutrients passed through to the body cells and from tissue cell to the blood. A network of capillaries joins the arteries and arterioles with venues and veins [4] [5].

1.2 Functionality of the circulatory system

The function of the heart is pumping the oxygenated blood from the lungs to various tissues of the body and from these tissues back to the lungs. This operation starts at heartbeat under the control of the Sinoatrial node (S – A node) or the Pacemaker, after receiving an electrical impulse from the brain in the right atrium. The SA node sends a signal to the A-V node (atrioventricular node) in the right ventricle and triggered to contract [3]. In this configuration, the left side receives the oxygenated blood and delivered into systemic arteries that progress this blood to the entire body. These form a tree of progressively smaller vessels that supply fully oxygenated blood to all organs and tissues of the body. From the smallest of the systemic arteries, blood flows into the systemic capillaries, which are roughly the diameter of a single red blood cell. It is in the capillaries that the actual exchange of oxygen and CO2 takes place. The blood leaves the systemic

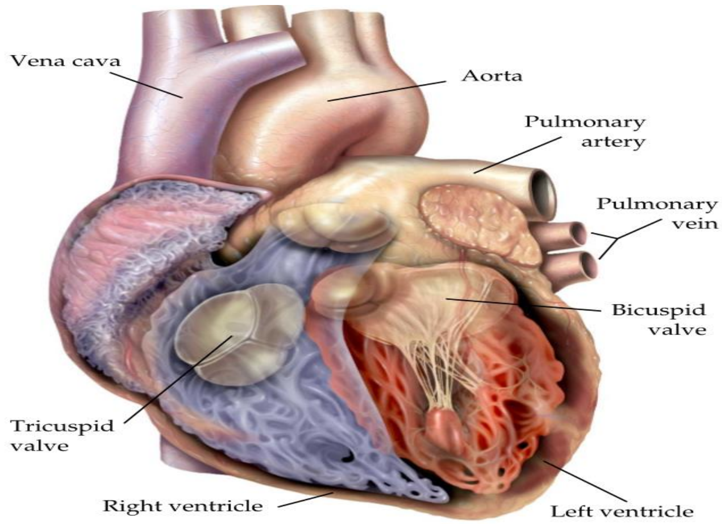


Figure 1-3 A diagram of the heart

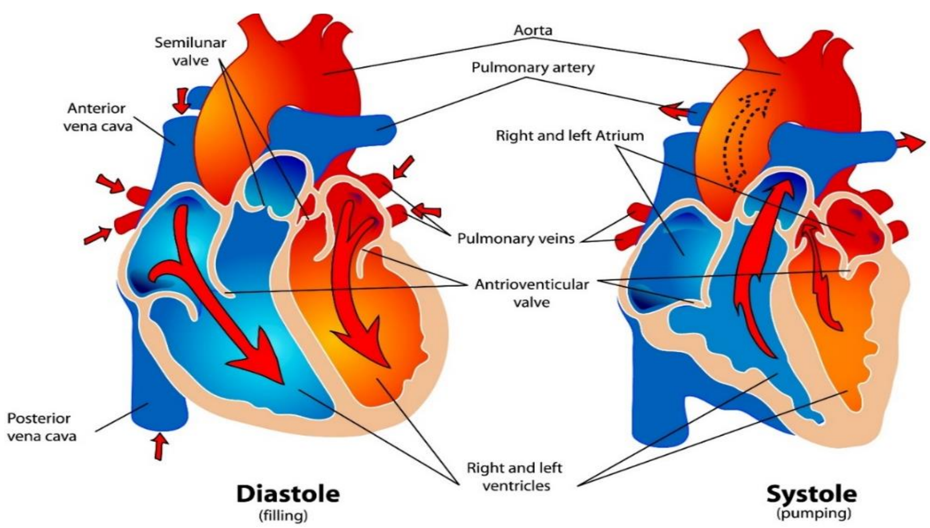


Figure 1-4 Cardiac cycle phases from Chapter 9: "Heart muscle: the heart as a pump and the function of heart valves"

capillaries with a smaller amount of O2 and more CO2 than the blood that entered. This blood enters the systemic veins and flows through the vessels of progressively increasing size toward the right side of the heart. The right heart pumps blood into the pulmonary arteries, which form a tree that distributes the blood to the tissues of the lung. The smallest branches of this tree give rise to the pulmonary capillaries, where CO2 leaves the bloodstream and O2 enters from the air space of the lungs. Leaving the pulmonary capillaries, the oxygenated blood is collected in the pulmonary veins, through which it flows back to the left heart [5].

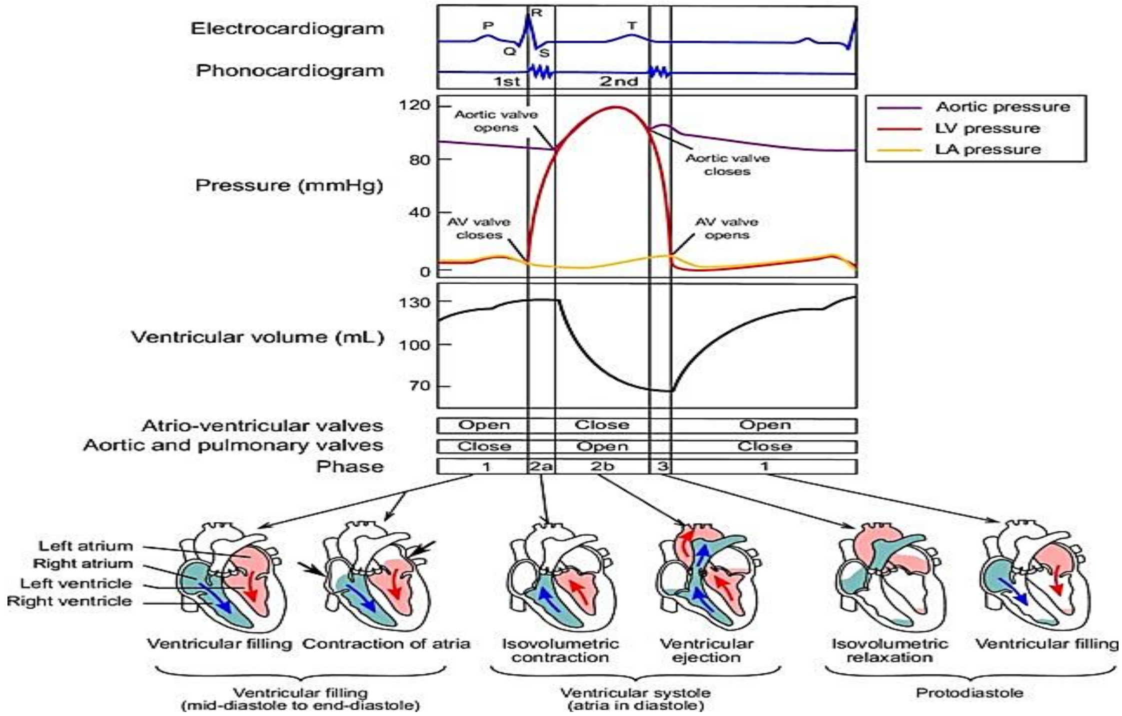


Figure 1-5: From (Marieb, 2005) Cardiac cycle: (top) diagram depicting cardiac signals (ECG, pressures, volumes) and cardiac events occurring in the heart. (bottom) illustration of blood circulation in heart cavities at each phase.

1.3 Cardiac cycle:

The cardiac cycle is the sequence of actions in one heartbeat. Simply, it is the simultaneous contraction of both atria, followed by the simultaneous contraction of both ventricles in a fraction of a second. The heart consists of cardiac muscle cells that connected to each other, and so when one contracts, they stimulate their neighbors and they all contract. The heart is getting its rest between beats. It can only respire aerobically. The heartbeat has two phases:

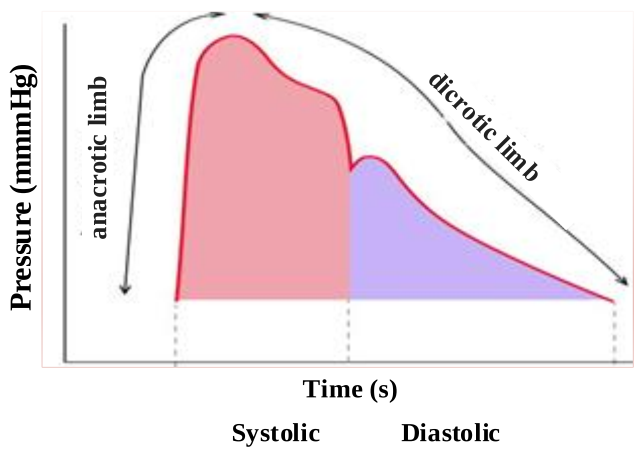
Systole: is the term for contraction. This occurs when the ventricles contract, closing the A-V valves and opening the Semilunar valves as shown in (Figure 1-4) to pump blood into the two major vessels leaving the heart.

Diastole: the term for relaxation. This occurs when the ventricles are relaxed, and allowing the blood backpressure to close the semilunar valves and opening the A-V valves [7]. (Figure 1-5) signifies the two phases occur in both sides in a different direction to permit the blood flow in the heart.

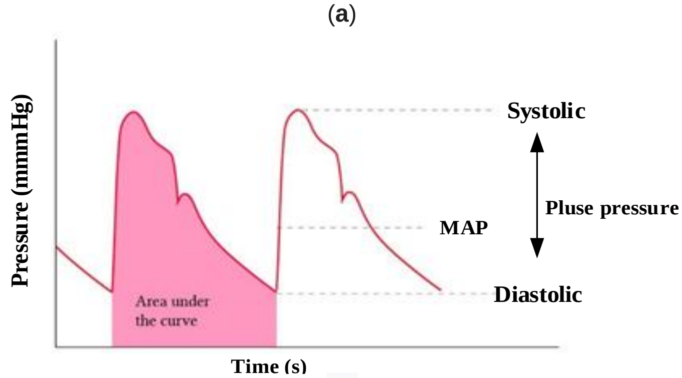
1.4 Pressure waveform

Arterial blood pressure waveform is a key measurement of the hemodynamic status, and it is a marker of adequate organ perfusion and tissue flow. It represents the impulse of the left ventricular contraction. The arterial pulse waveform can be separated into three components as shown in (Figure 1-6):

- The systolic phase is characterized by a rapid increase in pressure to a peak followed by a rapid decline. This phase begins with the opening of the aortic valve and corresponds to the left ventricular ejection as mentioned before.
- The dicrotic notch is a secondary upstroke in the descending part of a pulse tracing corresponding to the transient increase in aortic pressure upon closure of the aortic valve.
- The diastolic phase, which represents the flow of blood into the peripheral circulation.



(a)



(b)

Figure 1-6: (a) Pressure waveform represents systole (anacrotic limb) and diastole (dicrotic) phases.(b) pulse pressure with MAP mean value.

Some information can be derived from the arterial pressure waveform, regarding the change of the waveform depending on its position in the vascular tree [7].

For example, from arterial line amplitude can be defined:

- Heart rate
- Systolic pressure, Diastolic pressure (coronary filling).
- Mean arterial pressure (systemic perfusion)
- Pulse pressure (high in AR, low in cardiogenic shock), also the changes in amplitude associated with respiration (pulse pressure variation).

Besides, information from the arterial waveform shape some of them are:

- The slope of the anacrotic limb (the ascending portion of an arterial pulse tracing shown in Figure 1-6a) represents the aortic valve and L-V flow.
- A rapid systolic decline in the L-V valve.
- Bisferiens wave in Hypertrophic obstructive cardiomyopathy HOCM (Pulses Bisferiens, also known as biphasic pulse, is an aortic waveform with two peaks per cardiac cycle, a small one followed by a strong and broad one. It is a sign of problems with the aortic valve, including aortic stenosis and aortic regurgitation, as well as hypertrophic cardiomyopathy causing subaortic stenosis.
- Low dicrotic notch in states with poor peripheral resistance.

An ability to accurately and noninvasively estimate the pressure waves would indeed be a valuable addition to current non-invasive tools such as those provided by echo/Doppler and magnetic resonance imaging.

1.4.1 Pressure Volume relationship:

The relationship between pressure and volume in the ventricles is cooperative because the heart chambers are actively pressurizing and de-pressurizing over time, and thus the pressure-volume relationship is non-linear.

1.5 Blood Pressure

According to the United Kingdom (UK), heart organization, the blood pressure defined as the strength of pushing the blood during the vessel when the heart beats (which take around 0.8 seconds) [8]. Ideally, for the healthy people the blood pressure should be lower 120 over 80 mmHg (120/80), the top number is the systolic blood pressure, which is the highest pressure when the heart beats and pushes the blood around the body, and the bottom one is the diastolic blood

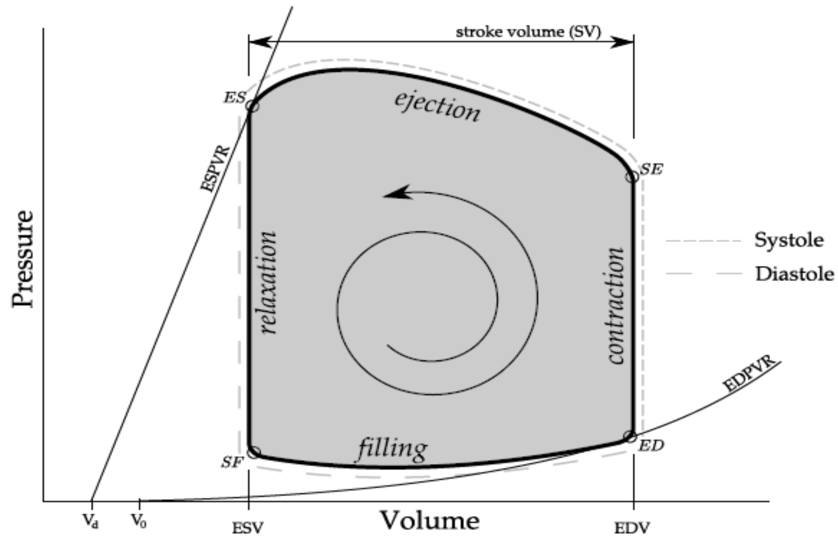


Figure 1-7: The pressure-volume loop where the subscripts ESPVR and EDPVR represent end-systole and end-diastole respectively

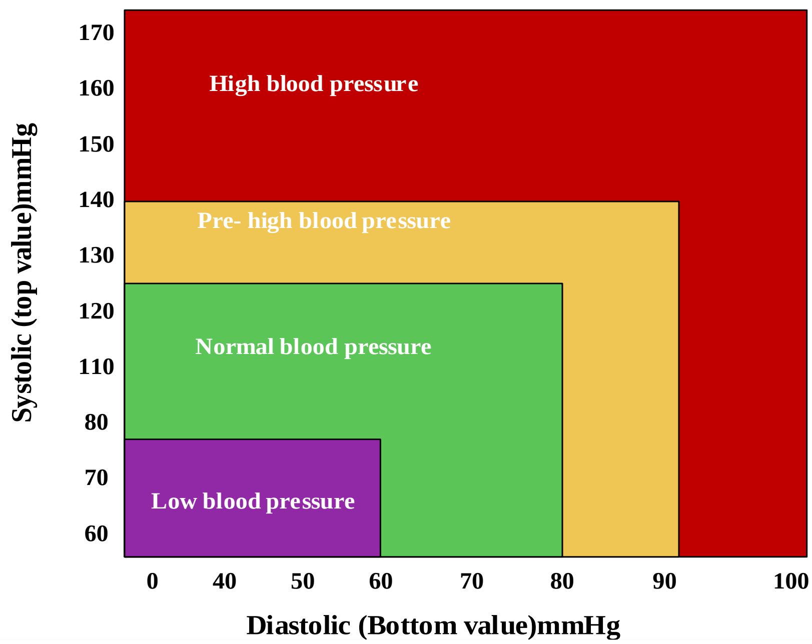
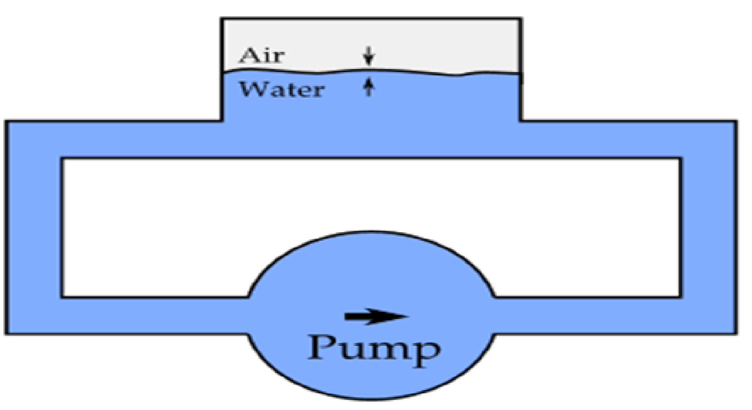
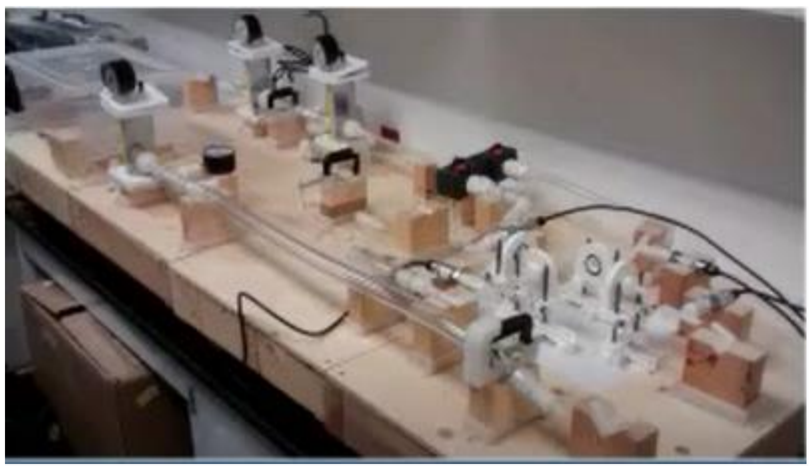


Figure 1-8 Block illustrate the blood pressure range

pressure that the lowest pressure when the heart relaxes between beats. Maintain the blood pressure at normal value is very important because high blood pressure is considered a risk on human health and causes problems in the future. High blood pressure is putting extra stress on the arteries and the heart; this strain can cause the arteries to become thicker and less flexible,



(a)



(b)

Figure 1-9 : (a) A hydraulic circuit consists a water pump connected to a chamber of air,(b) A mechanical model to the circulatory system

which leads to being narrowing and clogged, then cause a heart attack or stroke. The blood pressure illustrated in (Figure 1-8) below shows ranges of high, low and healthy blood pressure readings. Therefore, most the researches of modeling and representation consider this fact to evaluate the accuracy of their work.

1.6 Cardiovascular Representation:

Modeling and simulations represent a strong tool in the technical practice. It is widely utilized because of its effectiveness and variability. Applied to model the biological system allows studying

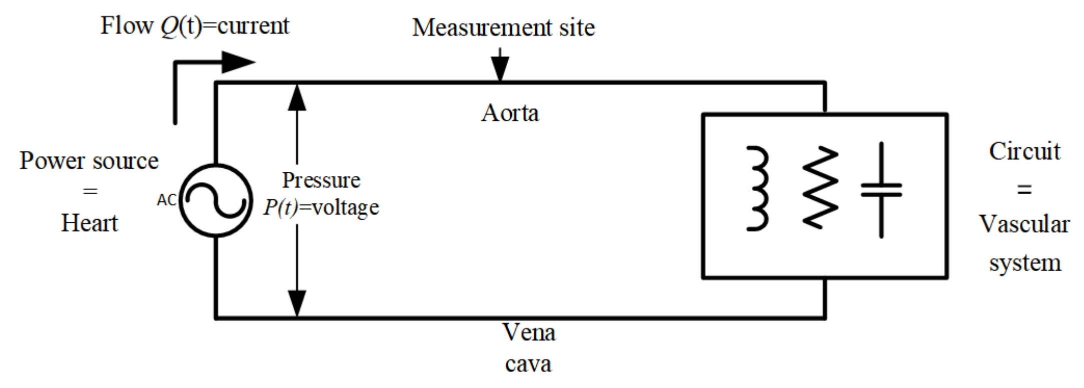


Figure 1-10 An electrical circuit describe the vascular system corresponding to analogy between the electrical current and systemic flow

a body under physiological conditions or evaluating the effect of pathologic changes in the organism and possible impact of therapy without any invasion. For a long time, several methods had been developed to discover the best approach that can describe the functionality of the cardiovascular system. Next, the mechanical and the electrical descriptions will provide an overview of the idea behind these models, and some examples will be re-simulated in chapter 2.

1.6.1 Mechanical Description

Some studies describe the heart as a mechanical pump and explain the heart rhythm by a mechanical expression. The most popular one was a Windkessel model and followed by numerous research. The Windkessel model was described by the German physiologist by Otto Frank (1899) [9], and since then has been used in some physiological models to simulate the elasticity of a compartment. The term Windkessel loosely translates from German as “air chamber”. The original model based on a hydraulic circuit consisting of a water pump connected to a chamber of air as shown in (Figure 1-9a). The water pumps to the chamber the water compresses the air that works to push the water out to return to the pump. This process establishes the equivalent to the movement of the blood on the body.

In (Figure 1-9 b), a recent mechanical model that currently used to verify the accuracy of the mechanical description to the real system. It performs an acceptable approximation, but it lacks the flexibility to change the tubes if the parameters were changed.

1.6.2 Electrical Description

Although the mechanical approach presents a good estimation, it is poorly defining some tricky blood flow characteristics in the vessels. Studies move further and establishing an electrical circuit model using a lumped element to achieve a description of the necessary system properties and dynamics with a minimal amount of complexity.

The (Figure 1-10) expresses the analogy between the blood flow in the systemic arteries and the current in an electric set of mathematical equations between some of the variables of the system, mainly blood pressures and flows that describe these models.

These variables interrelate by a set of parameters (blood viscosity, vessel diameter, vessel wall elasticity, etc. The next table shows the most common variables between the two systems, and its equivalence in physiology [10], chapter 2 will provide extra explanations about this analogy.

Table 1-1: Analogy between the fluid dynamic, hemodynamic and electrical variables

Fluid dynamics	Physiology. Variables	Electrical analogue
Pressure P [Pa=J/m^3]	Blood pressure. [mmHg]	Voltage *U* [V=J/C]
Flow rate Q[m^3/s]	Blood flow rate[L/s]	Current *I* [A=C/s]
Volume V [m^3]	Blood volume[L]	Charge q[c]
Viscosity µ	Bl.res. $R = \frac{8\ \mu N l}{\pi r^4}$	Electrical resistance R
Elastic coefficient	Vessel's wall compliance	Capacitor's C
Inertance	Blood inertia	Inductor's Inertance L
Poiseuille's low	$Q = \frac{\Delta P}{R} = \frac{\Delta P \pi r^4}{8 \mu l}$	Ohm's low: I=$\frac{\Delta U}{R}$

(Figure 1-11) shows the electrical illustration of the cardiovascular system in Pater [11] [12]. His analogy still used in some recent researches and provides an acceptable representing of the cardiovascular system. On the other hand, the complexity to build this model eliminates mobility due to the size. This bulky and expensive analogy is a non-practical solution.

The next chapter will examine some models that use the electrical representing and how they have estimated the parameters

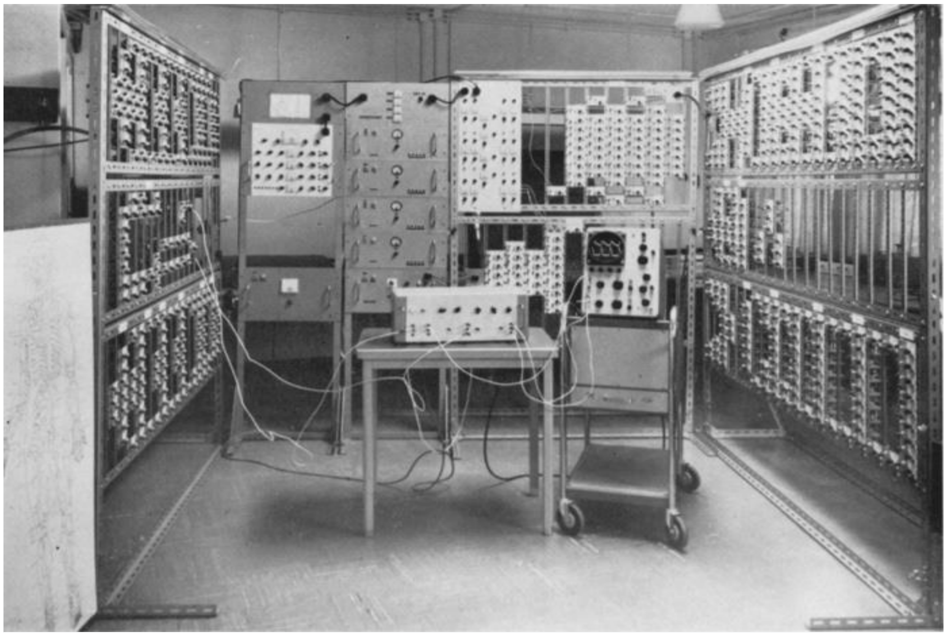


Figure 1-11: An electrical illustration of the cardiovascular system of the human body (1966).

Chapter 2 : Lumped Element Cardiovascular Modeling

This chapter offers an overview of the history of circulatory system concepts. The most important parameters will be assigned to represent the quantitative description of the circulation. In addition, providing a review of the previous models that have been done and re-simulating using ADS (Advanced design system), and defines their limitations.

2.1 History of Cardiovascular Modeling (Lumped element)

The prior consideration of blood circulation was that the food is converted into blood in the liver to produce the energy, but in 13th century, Ibn Al-Nafis (1213-1288) was the first physician who was correctly contributed the pulmonary circulation [13] to against the wrong common thought. Later, in 1628, William Harvey (1578-1657) demonstrated experimentally that the blood is pumped from the heart and circulated in the body. In addition, he made the first publication on this consideration [14]. After that, in 1738, Daniel Bernoulli (1700-1782) investigated the laws governing the blood pressure and published his well-known Bernoulli's equation, which relates the blood pressure to the blood velocity [14]. In 1775, Leonhart Euler (1707-1783) introduced the first model of pulse wave propagation for in viscid fluid [15] Next, Thomas Young (1773-1829) presented a mathematical model describing the blood flow. In 1838, Jean Lonard Marie Poiseuille (1797-1869) and in 1839, Gotthilf Heinrich Ludwig Hagen (1797-1884), independently derived a physical law that explained the relationship between the pressure drop and blood flow under steady flow conditions [16].

Later, Poiseuille published the physical law, known as Hagen-Poiseuille law in 1840 and 1846. In the field of cardiac physiology, Otto Frank (1865-1944) made several important contributions, such as "Fundamental form of arterial pulse", which was the first theory of" Windkessel effect" that was detected by Hales in 1733 in the field of blood circulation[17].

However, (1963) Noordagraaf [18],(1966)Pater[11][12],(1968) Westerhof [19] [and (1980) Avolio [20] used the concept given by O. Frank (Windkessel theory) and constructed the electrical analog model of the major arteries in the systemic circulation, an electrical circuit, composed of a resistor, inductor and capacitor, represented each segment. From the last few decades, lumped-parameter models of cardiovascular system have gained attention to study various normal and pathological conditions [21]-[23] in (2007) (2009) (2011). Lumped-parameter models are simple, computationally less expensive and explain the physics of the real-world problems well enough.

2.2 Parameters of the existing models

The purpose of this section is to introduce some essential physical variables in a quantitative description of the circulation, and its similarity to the electrical quantities.

2.2.1 Flow, Volume, and Pressure

Firstly, the heart beating is the responsible of the blood movement, because of the pressure difference, similarly, a battery (or power supply) develops a voltage "Potential Difference". Thus, a fluid pressure drops (energy per unit volume) corresponds to a potential difference, or "Voltage drop" (energy per unit charge)[23].

Pressure (P) =Voltage (V)

mmHg = Volt

Likewise, the blood flow is the volume of blood per unit time passing a point in the circulation, the flow is mostly measured by liters/minute, which the same as circuit's electric current which is the rate of movement of charge.

Flow (Q) =Current (I)

Liter/min = Ampere

Applying Ohms low is very close to Poiseuille is low by replacing the pressure drop to voltage;

$$\mathrm{R} = \frac{\mathrm{P1} - \mathrm{P2}}{\mathrm{Q}} \tag{2.1}$$

$$\mathrm{R} = \frac{\mathrm{Voltage(V)}}{\mathrm{Current(A)}} \tag{2.2}$$

2.2.2 Capacitors and RC- Circuits:

The estimation of the electrical / fluid analogy capacitance is more complicated than the resistance. As a stored component, so the sameness is the arterial walls which stores the energy from the pulsatile flow [24], therefore "capacitance" of a capacitor is stated in terms of the amount of charge (Q) stored at a given voltage drop (Across the capacitor).

$$C = \frac{Q}{V} \tag{2.3}$$

Obviously, after this quick preview of the analogy, the electrical lows can be applied. (Table 2-1) represents the correspondence between the cardiovascular with its relevant electrical quantity.

2.3 Lumped element models:

Lumped-parameter models are used to estimate the total arterial properties by fitting the model to the measured (aortic) pressure and flow. A mathematical model of a physical system, which is simplified from spatially distributed variables in a single scale variable, is called a lumped parameter model. Lumped parameter models are in common use for studying the factors that affect pressure and flow waveforms.

Table 2-1 The analogy between the electrical and cardiovascular fluid

Cardiovascular system	**Electrical symbols**	**Electrical equation**	**Cardiovascular fluid equation**
Vessel Resistance (Rc)	Resistance (Re)	V = I. Re V = voltage I = current Re= circuit's Resistance	P = F. Rc P = pressure F = blood Flow Rc= vessel's Resistance
Vessel compliance (Cc)	Capacitance (Ce)	Ce.$\frac{dv}{dt}$ = I Ce= Circuit Capacitor	Cc.$\frac{dp}{dt}$ =F Cc= Vessel's compliance
Blood inertia (Lc)	Inductance (Le)	Le. $\frac{dI}{dt}$ =V	Lc $\frac{dF}{dt}$ = P

2.3.1 Current source (Flow) analysis and simulation:

It is important to re-simulate the adequate input as was used for the models. The real input for the heart; is the left ventricle blood flow, which is delivered per stroke as a function of time with a

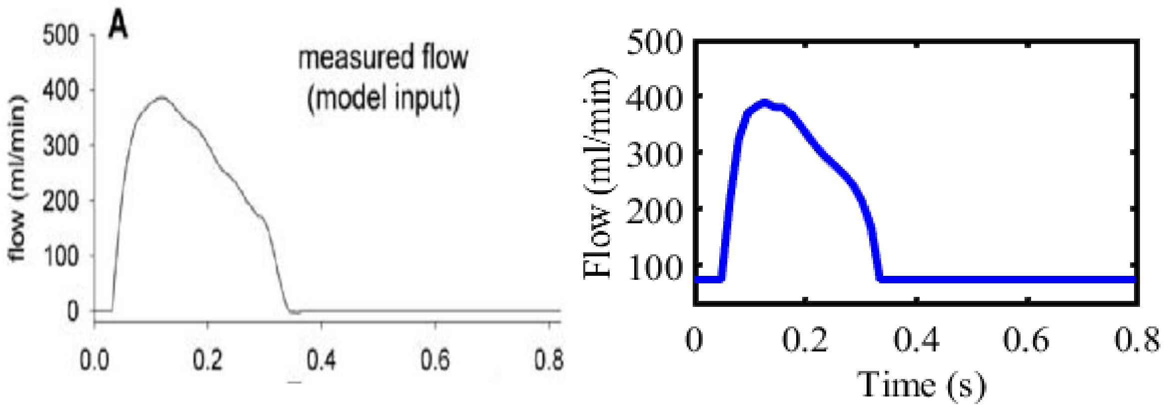


Figure 2-1 a) Measured flow input from [25] with *f*=1.29 Hz (equivalent to 77 beats/minute) b) ADS simulation of the flow input. The ADS output current originally with Ampere as unit is illustrated in the figure as Flow (ml/min)

frequent beat 60 over 70 per second around 0.8 second. Based on experimental measurement signal curve shown on [25] illustrate in (Figure 2-1a), the real data curve was generated under some conditions and ***f***=1.29 Hz in ADS (equivalent to 77 beats/minute) as seen in (Figure 2-1b).

Different model configurations are coexist, here some lumped element models will be tested and simulated using ADS (Advanced Design System) software to demonstrate the resulting pressure and illustrate their accuracy.

2.3.2 The three-element Windkessel (WK-3) model:

The first model that was tested is the most popular one that had been simulated in many researches and mainly consisting of R1, C, R2 (vascular resistance R1, total arterial compliance C, and characteristic impedance R2). As a referenced in [25]-[28], the measured flow is used as an input to the Windkessel model. The used lumped-parameter values were optimized during the result curve fitting that was used and the same values are used in ADS simulation. In this model,

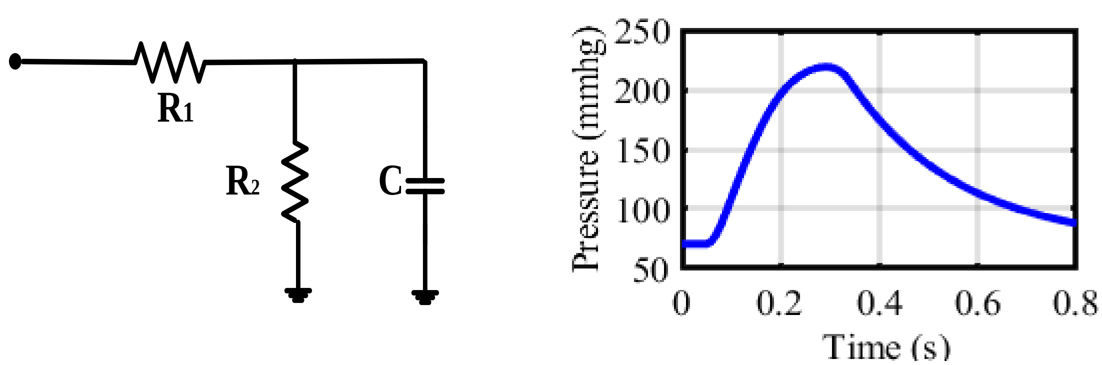


Figure 2-2 Three elements Windkessel model and its result by ADS.

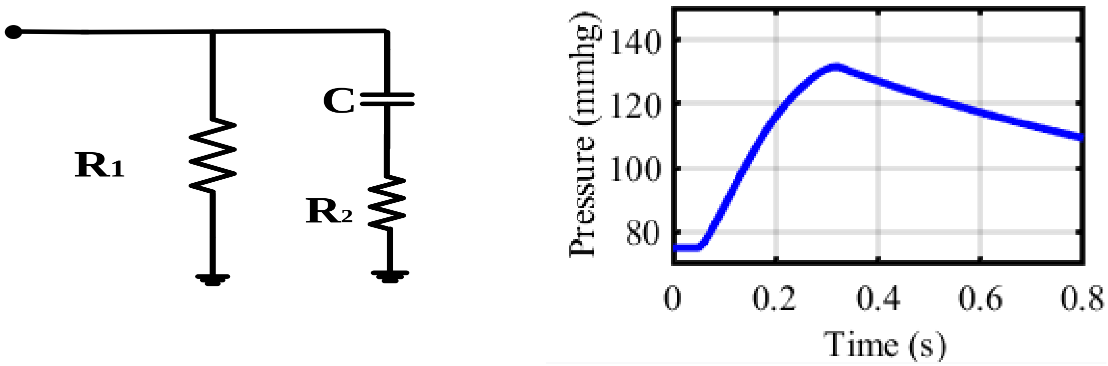


Figure 2-3 Three element with series resistance in the load.

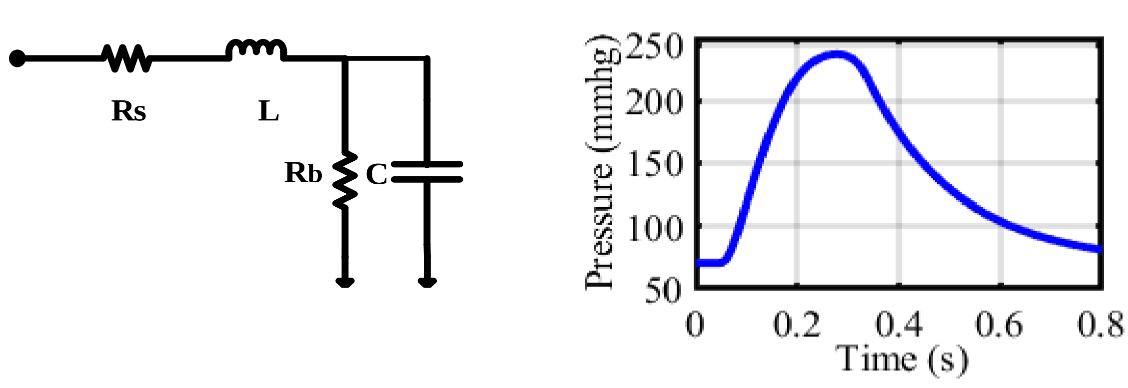


Figure 2-4 Four elements Windkessel model and its result.

the input impedance represents by resistor as a consideration of the ratio of mean pressure over mean flow, and peripheral resistance and arterial compliance are the load.

Although the resulting curve is close to the aortic pressure, but the resulting pressure is higher than the normal value. Therefore, the idea behind the approximated fitting is not effective beside this model limit to account the elastic of the vessel, therefore some researches were derived more complex model to improve the fitting results of these models and add more consideration to other blood vessel characteristics.

2.3.3 Four –element Windkessel (WK-4)

The Windkessel four elements model has been proposed to reduce the error that was occurred by the three-element model, a component accounting for the inertia of the blood, L, provides an

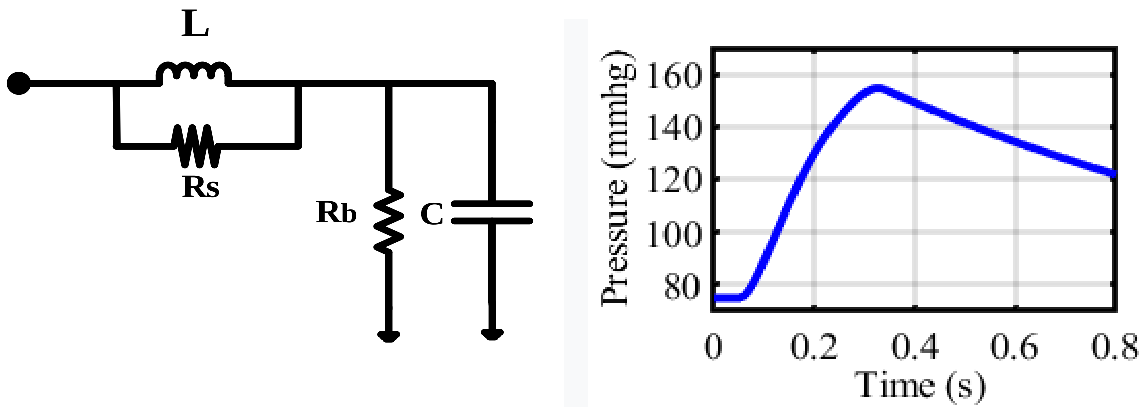


Figure 2-5 Four-element model with input parallel inductance and resistance

improvement over the three-element Windkessel models. Thus, consists of four elements: R1, C, R2, and however, there are two different versions of this model: one with the inductance element placed in series with R1 (WK4-s) and the other one will be in parallel with R_1 [25].

In this model (Figure 2-4) suggested to add an extra element to the input impedance. This inductance implies an increase in the impedance.

(Figure 2-5) also represented by four elements but different configuration, here the inductance parallel with R_1.

As shown in (Figure 2-4) and (Figure 2-5) it seems that the 4-element model gives a more accurate representation of the blood pressure versus the cardiac cycle time curve when compared to the three element models.

Other models using lumped element, and their parameters dependent on certain equations to evaluate each element value. These equations depend on vascular characteristics and their equivalence to Poiseuille’s lows.

Some concepts about the blood and fluid dynamic and blood vessels properties should be introduced to understand these equations. Based on the symmetry between the circulatory system and hemodynamic so Poiseuille’s equations can be applied through the pressure, viscosity, flow rate, velocity of blood, and vessel diameter. The viscosity (μ) defined as the

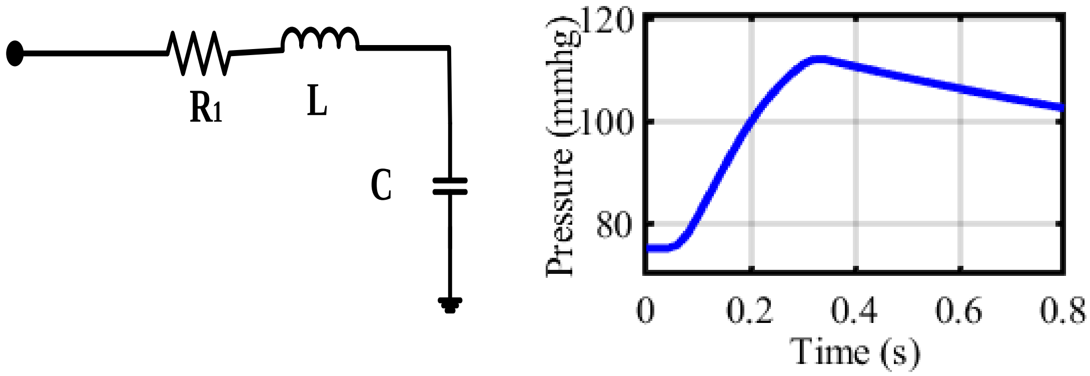


Figure 2-6 Three element model applying equations.

resistance of fluids against flow, so particularly for the blood is the resistance to the blood circulation includes the blood features and the vessels shape [29]. The fluid flow as it customary known needs to an energy to drive through a circulation, therefore, the energy of the circulatory comes from the pressure difference [30]. Density is defined as mass per unit volume. The blood is a mixed liquid and its mass contains 55% plasma and 45% cells (red, white, and platelet), the average density of whole blood for a human is about (1050-1060) kg/m3 [30]. The density reduction can be used for the determination of distribution volumes, of flow through organs, and of the cardiac output by measuring the velocity of transmission. Beside the blood characteristics, the vessel mechanical properties also have the effect on the blood flow and pressure, Elasticity property of the vessels affect the behavior and function of the dynamic characteristics of the circulatory system as well as the wall thickness of the blood vessels. Here the blood velocity can be defined, as the rate of blood flow through a given vessel, which is inversely proportional to the cross-sectional area of the blood vessel. It is important to know that changing in mechanical properties of arteries have been associated with various diseases.

Now the model in [31] [32] use 3-elements C, L, R to represent each vessel, aorta dimension are apply to calculate them

$$R = \frac{8.\pi.l.\mu}{A^2} \tag{2.4}$$

Where µ is blood viscosity, l and A are in respect length and cross section area of each artery segment.

$$L = \frac{9.l.\rho}{4.A} \tag{2.5}$$

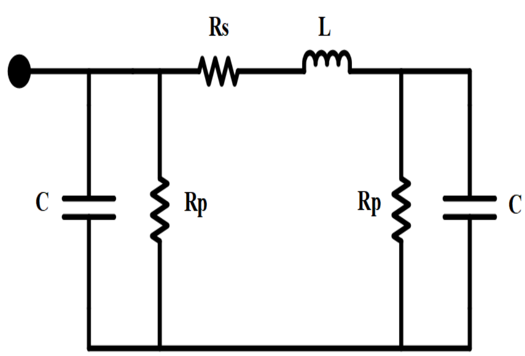
Where ρ is blood density.

$$C = \frac{3.l.\pi.r^3}{2.E.h} \tag{2.6}$$

Where r, E, and h are in respect artery radius, Elasticity module and thickness of arteries. The results of simulation performed for a heart frequency of 1.219Hz

In the equivalent circuit, aortic output is acceptable (Figure 2-6), but it can be more accurate by adding another element.

Other approaches use other equations by π-Filter model (Figure 2-7). Principally, the model is based on an electrical analogue of the hepatic blood flow through a dog liver like any electrical analog model, it is based on the analogy of the simplified Naiver-Stokes equations describing the flow through a cylindrical tube (representing a blood vessel), in this case a π-filter [33]. The filter contains four different components defined in (2.7) – (2.10): serial resistance R_s, serial inductance L, parallel conductance C, and parallel resistance R_p. In the original work, the R_p and capacitance



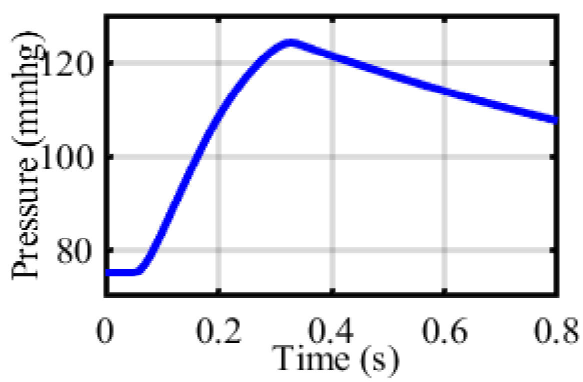


Figure 2-7 π -filter model

C are in series. The result shown in (Figure 2-7), illustrate that the modification to parallel achieve excellent results.

$$R_S = \frac{8.\mu.l.}{\pi r^4} \tag{2.7}$$

$$L = \frac{1.33. \rho . l}{\pi r^4} \tag{2.8}$$

$$C = \frac{l^2}{2\ cp^2 L} \tag{2.9}$$

$$R_P = \frac{2 * 10^{-6}}{\mathrm{C}} \tag{2.10}$$

Where c_p is pulse wave velocity, r is the radius of the vessel.

The result of the last model is more accurate and closer to real measurement however that the effect of the extra capacitor and resistance can be neglected, and there is no big difference, if just simulating with load one capacitor and resistor.

2.4 Conclusion

Lumped element models that represent the cardiovascular system provide a good understanding of cardiovascular function. However, to improve the accuracy of these models, they should be specific when defining the blood characteristics and the vessels dimensions. The previous models have some limitations as (WK-3) and (WK-4) because some blood characteristics and flow distribution cannot be studied or recognized. Therefore, models with equations are more acceptable, they can calculate the parameters if the radius and length changed, and they respect the elasticity module and thickness of arteries. The π -filter model provides a reasonable result; however, the effect of extra R-C can be ignored. It considers useless if the model was just with R, L, with parallel C, R, the model will be as in (Figure 2-4) and the result will be the same. This configuration is similar to the transmission line lumped element representing. If the transmission line lumped element representing is cascaded, so it will represent the π-filter model as (Figure 2-8) . The next chapter will present more details in this concept.

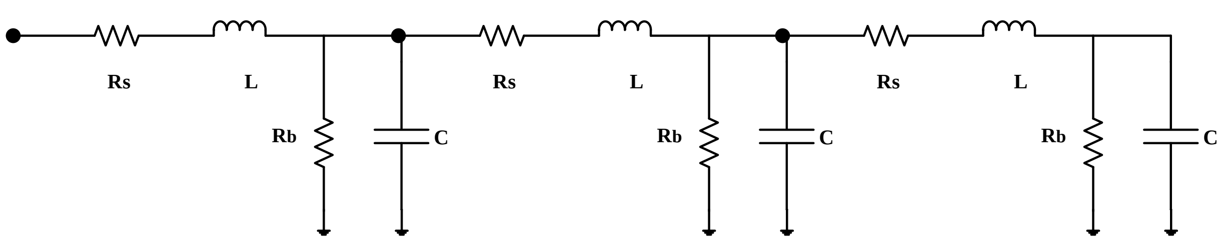


Figure 2-8 Cascade to four-element model to represent π-filter model

Chapter 3 : Frequency Scalability

3.1 Introduction

Lumped models for blood flow and pressure in the systemic arteries were presented in chapter two. Some models depending on fitting or estimation, other models on equations. In this chapter, the effect of increasing the frequency on lumped element will be studied to demonstrate if the same result can be reached at RF frequency. As offered in the previous chapter, the analogy between the hemodynamic and the electrical was representing the models that describe the circulatory system.

In Pater [11] thesis, he started to demonstrate the analogy between the hemodynamic system and transmission line theory and derived equations related to the voltage and the current. His representation was made by π-filter and he considered each length *l* of the segment of the transmission line could represent by one π-filter.

The proposed model following these forms which was used in chapter two in π-filter model;

$$R_S = \frac{8.\mu.}{\pi r^4} * l \tag{3.1}$$

$$L = \frac{\rho}{\pi r^2} * l \tag{3.2}$$

$$C = \frac{3\pi r^3}{2Eh} * \ l/2 \tag{3.3}$$

$$R_P = \frac{2 * \mu\, h}{3\pi r^3} * 2/l \tag{3.4}$$

The accuracy of these equations depends on the ratio length of the segment to the wavelength which is l = λ/8. The estimation that was used in his contribution mainly focused on the heart frequency, but also it was examined at higher frequencies and a significant error was found in the pressure waveform damping because of the different long length that needed at higher frequency. The availability of an accurate software as ADS give us the opportunity of estimated more adequate rules. On other hand, the principal idea behind the analogy could demonstrate better insight corresponded to this theory. In next section, the validity of increasing the frequency has been studied with a specific coherent process.

3.1.1 Scalable model with frequency

Lumped parameter models are useful to study the relationship of the Cardiac output to different parts in the human body that can be referred to them as (peripheral loads),but because of the finite number of lumped elements, they cannot model the specific determination of the system without adding many more elements.

Firstly, to starting the initial circuit, a Sine source at 1 Hz is used, depending on a corresponding blood flow into the aorta from the ventricle during the cardiac cycle as shown in (Figure 3-1). Previously in [35] was modeled as a sine wave with amplitude I_0 during systole and is zero otherwise and this follows the cardiac physiology. During diastole, when the ventricles are relaxed, there is no blood flow into the aorta, and therefore, $I(t) = 0$. However, with ventricular contraction during the systole, blood is ejected into the aorta and can be modeled as a sinusoidal wave, therefore:

$$I(t) = I_0(\sin(\pi * \frac{\mathrm{mod}(t,\mathrm{Tc})}{\mathrm{Ts}})) \tag{3.5}$$

Where t is time in seconds, Tc is the period of the cardiac cycle in seconds, Ts is the period of systole, in seconds, and mod (t, Tc) represents the remainder of t divided by Tc. Ts is assumed to be 2/5 Tc, according to the dynamics of the cardiac cycle. According to literature, the blood flow in one cardiac cycle is 90 cm^3. So, theI_0, was obtained as fallow:

$$90 = \int_0^{TC} I_0(\sin\left(\pi * \frac{\mathrm{mod}(t,\mathrm{Tc})}{\mathrm{Ts}}\right) dt \tag{3.6}$$

$$I_0 = \int_0^{TC} 1/90(\sin\left(\pi * \frac{\mathrm{mod}(t,\mathrm{Tc})}{\mathrm{Ts}}\right) dt \tag{3.7}$$

I_0= 424 .1 mA

Therefore, the maximum amplitude of the blood flow during systole is I_0 = 424 .1 ml. The input source was controlled under these principles.

Comparing the lumped element equivalent circuit of transmission line with the four-element Windkessel [25] model that was previously simulated. In this simulation, the result was approximate by fitting the curve to the measured curve. However, as shown in (Figure 3-2), the pressure is higher than normal range.

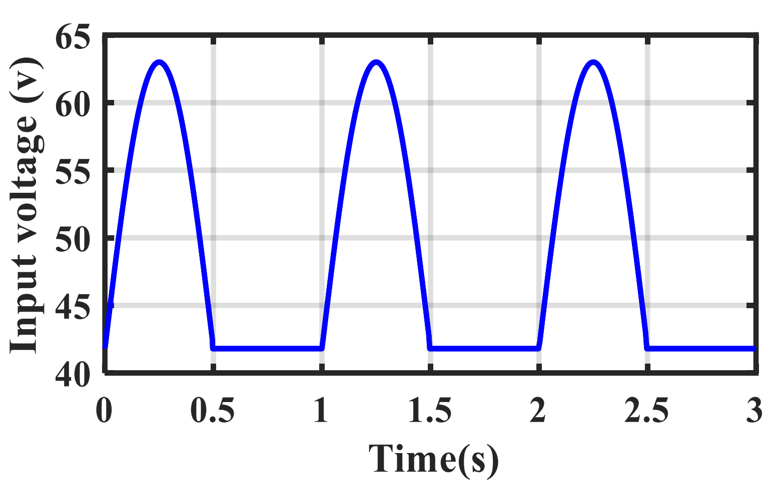


Figure 3-1 Input signal to transmission line lumped element circuit

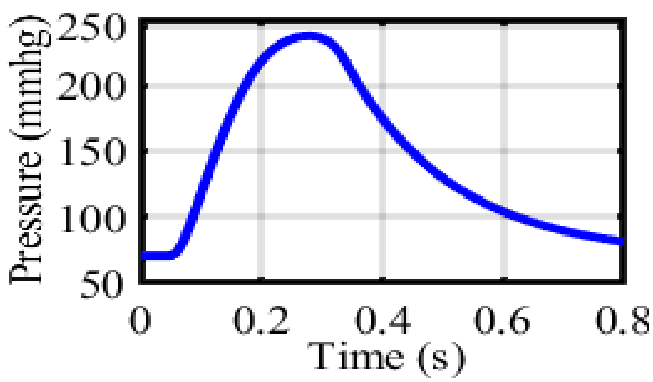


Figure 3-2 Four element model result

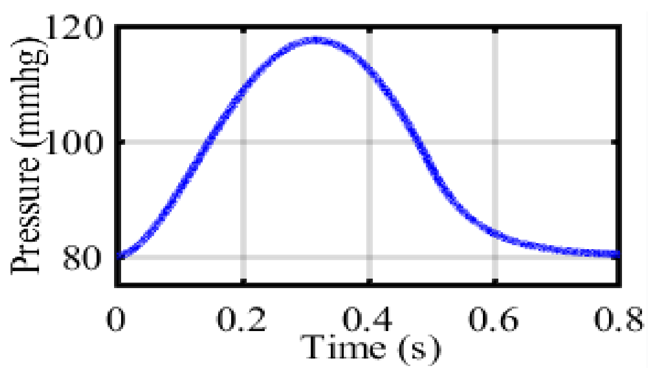
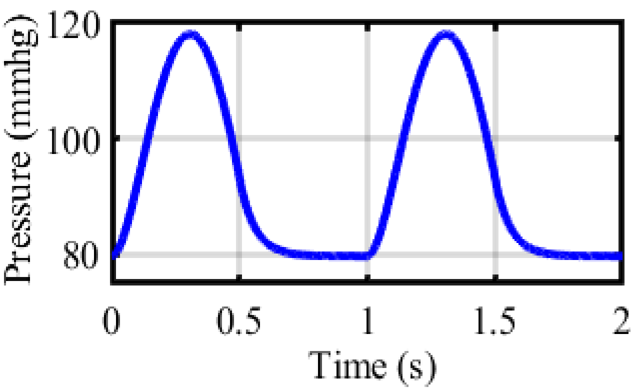
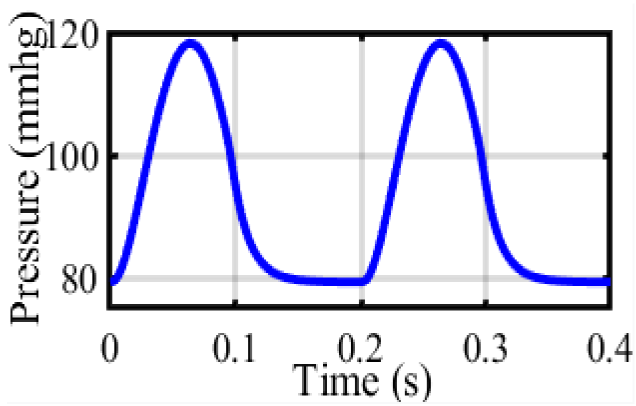


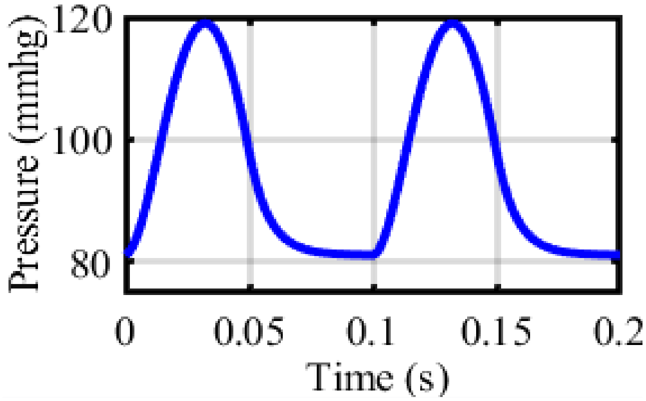
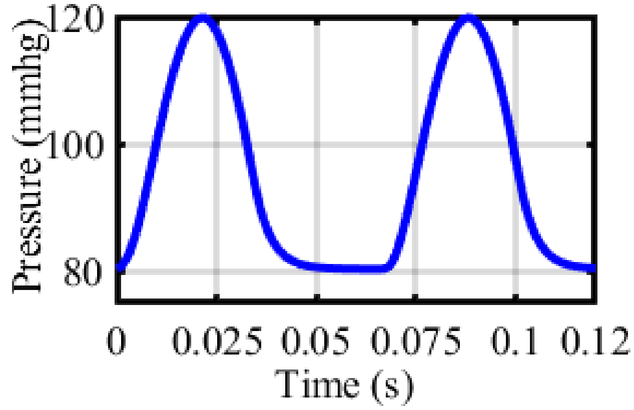
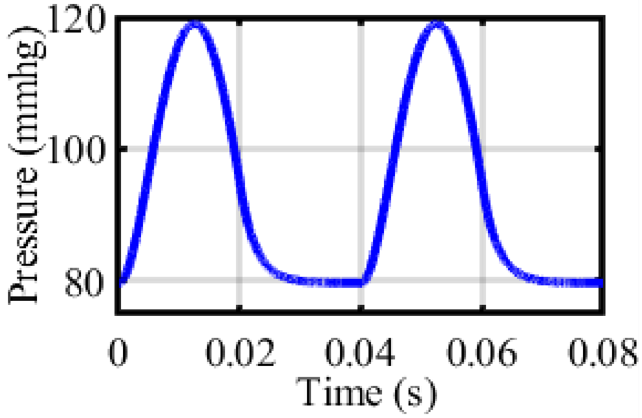
Figure 3-3 Transmission line lumped element output at 1Hz

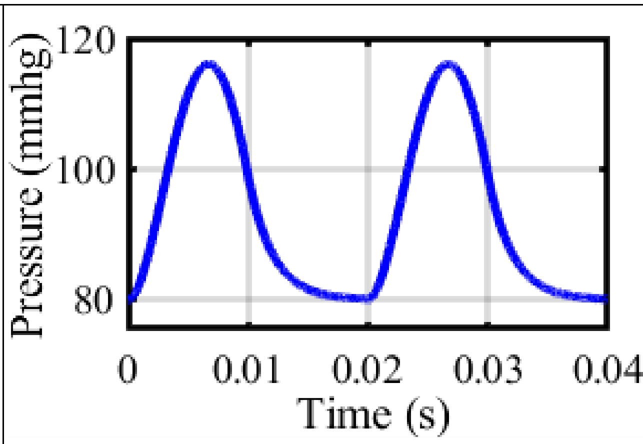
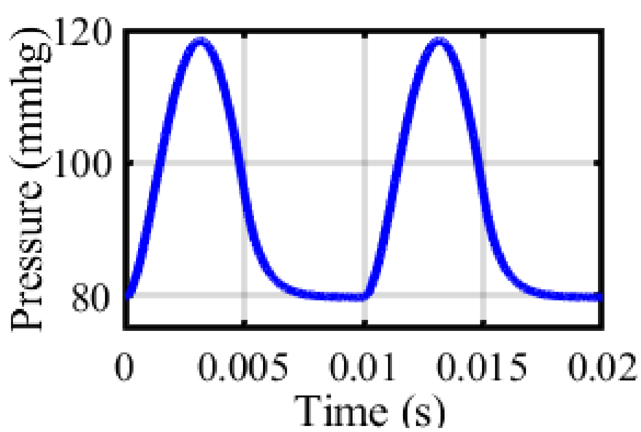
Thus, the parameters should be modified to have the typical value of the pressure. Incidentally, in the initial circuit the value of the elements was calculated by the previous equations with one-segment aortic dimensions as referenced in [31]. (Figure 3-3) shows the simulation result for the model at frequency of ***f*** =1 Hz. The equations and the circuit provide an acceptable result. To examine if this circuit can work at higher frequencies, the number of cells will be cascaded identically, as the frequency is increased as well. Applying tuning feature on lumped elements parameters until the output reach the normal range pressure result to study the influence of the frequency on them.

This simulation was done from 1Hz to 100Hz and each time, the values of circuit's parameters are conducted to estimate the effect of increasing frequency on each element (Table 3-1) illustrates the quantity of each lumped element at each frequency.

Table 3-1 : Signal output at each frequency and the value of each element.

Frequency (Hz)	Output pressure (volt –mmHg)	L (H)	C (pF)	R_1 (K-Ω)	R_2 (K-Ω)
1		3.07e4	6.8	16.7	300
5		1.3e3	6.8	0.5	540

10		307	6.8	0.158	690
15		137	6.8	0.075	818.4
25		52	6.8	0.021	966.2

50		13.8	6.8	0.007	1162.1
100		2.89	6.8	0.0017	1382.2

As shown in the previous table, each time that the frequency is amplified the L, R1, R2 are affected somehow by changing in the frequency. As noticed the inductance L and R1 are decreased in other hand the C became constant, but R2 is increased differently by increasing the ***f***.

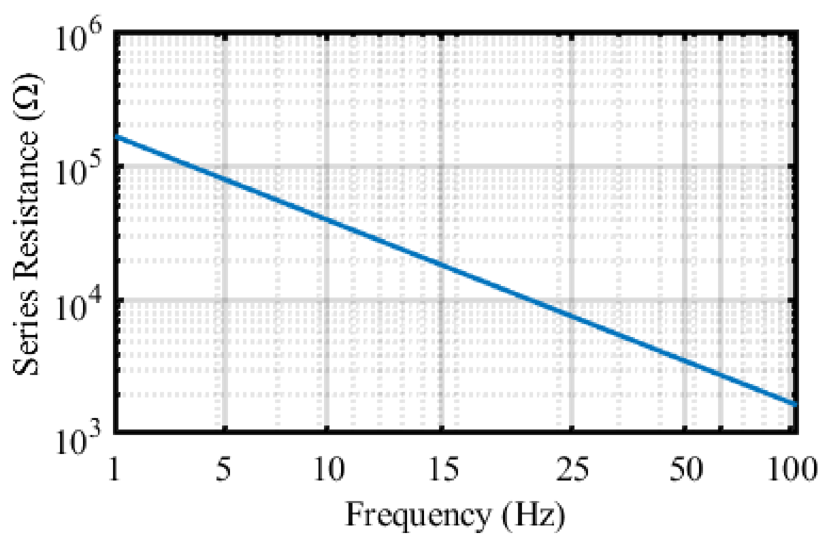


Figure 3-4 Serial resistance with frequency.

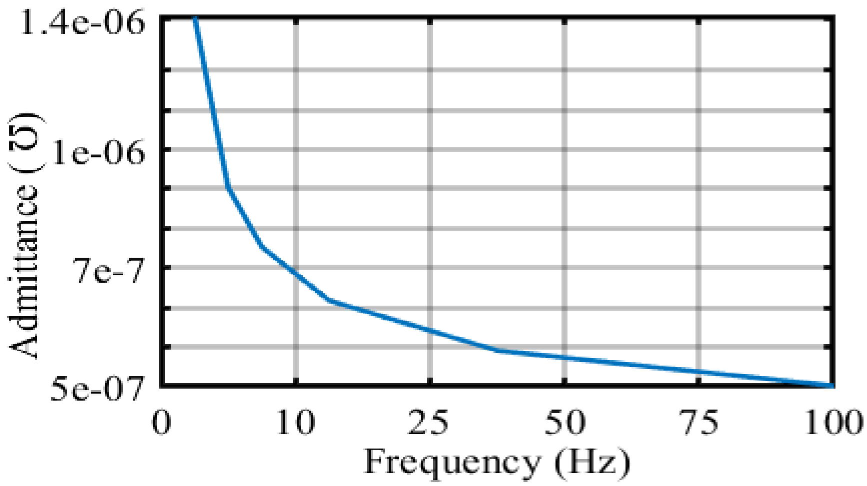


Figure 3-5 Parallel admtance with frequency.

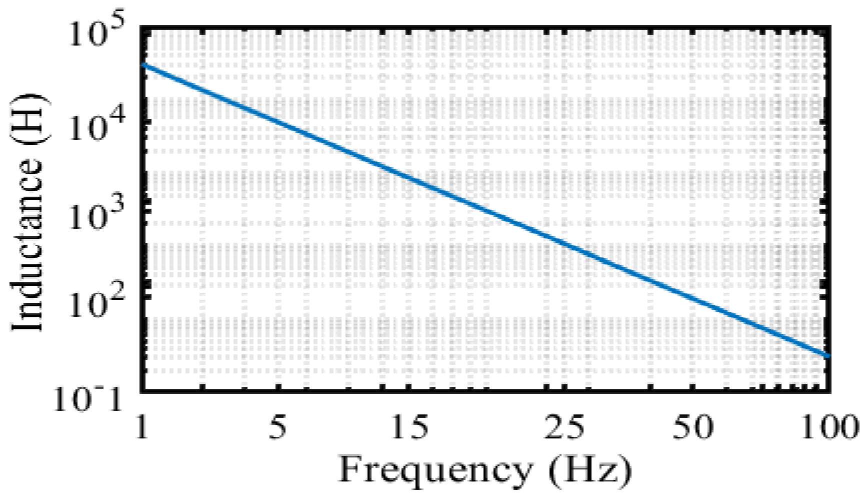


Figure 3-6 Inductance with frequency.

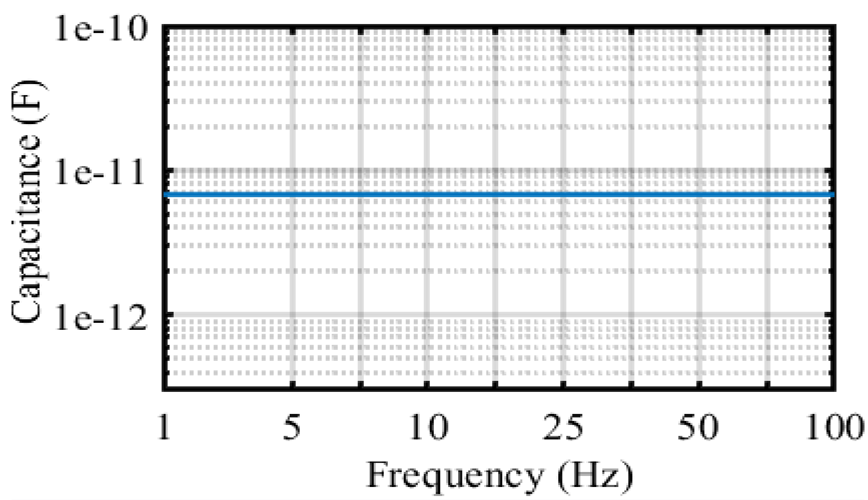


Figure 3-7 Capacitor with frequency.

In order to continue with higher frequencies, number of cells will impede the process. The following expressions will describe each element dependency to the frequency change as shown in the (Figure 3-6), and (Figure 3-7) by following formulas:

$$R(f) = \frac{\mathrm{R}_1}{f^2} \tag{3.8}$$

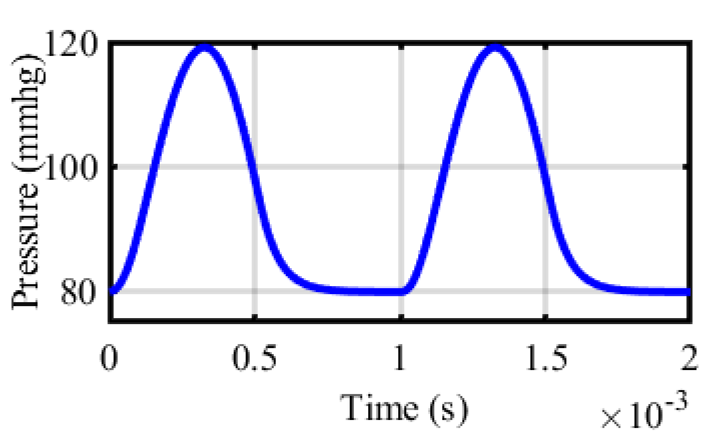
$$Y(f) = Y * \ln(f) \tag{3.9}$$

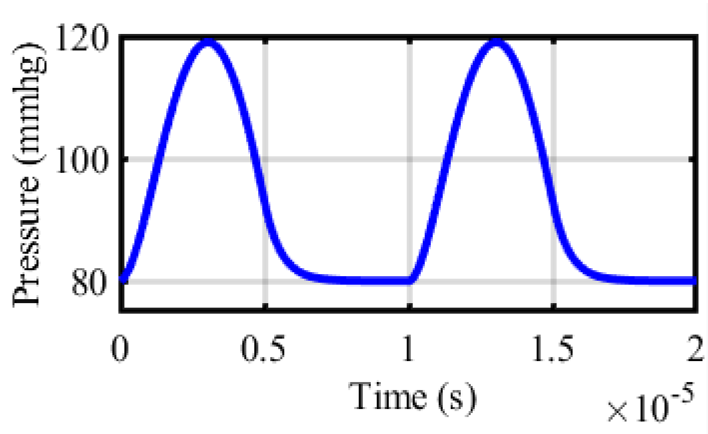
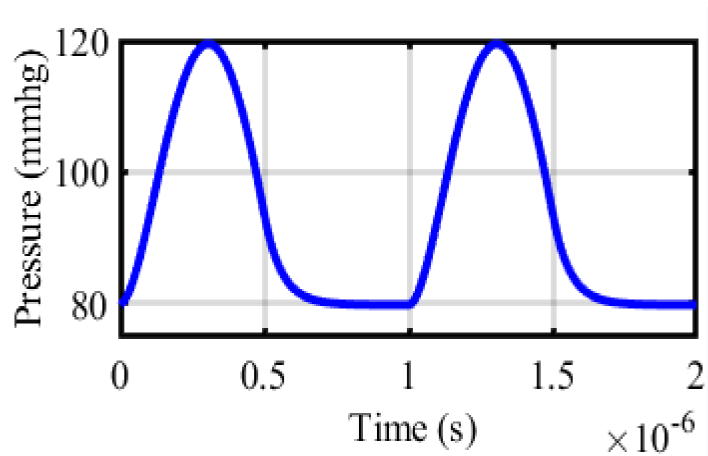
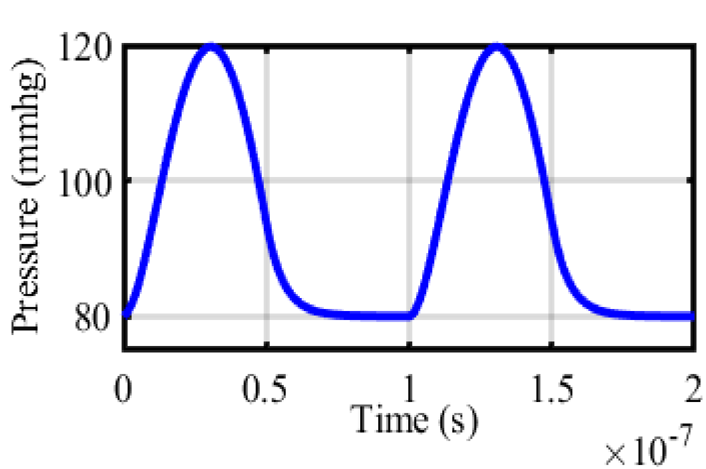
$$L(f) = \frac{L_1}{f^2} \tag{3.10}$$

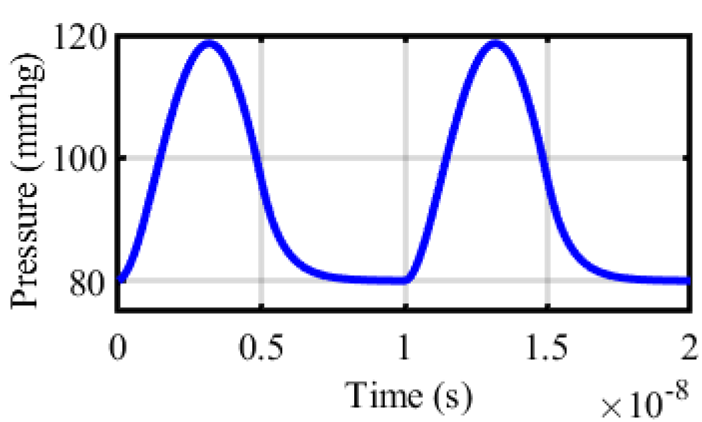
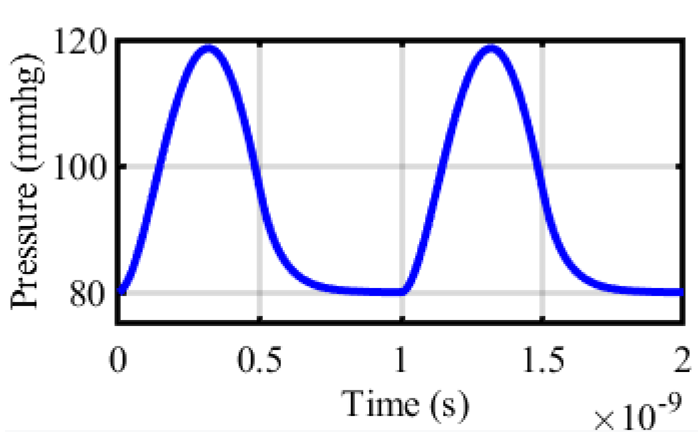
$$C(f) = C1 \tag{3.11}$$

These equations demonstrate the relation between each lumped element and the frequency *f*. In addition, the influence of these parameters on the waveform during the tuning was notified, which is provide an important information to define the changes if it is occurred, this will be discussed in future work. Now higher frequencies can be easily simulated using one cell. Next table will show the output at each frequency and at the same time will validate our equations.

Table 3-2 : The output pressure at high frequency

Frequency	output
1KHz	

100KHz	
1MHz	
10MHz	

100MHz	
1GHz	

As noticed in (Table 3-2), each time the frequency increases the transient increase with the same rhythm with different scalable time. (Figure 3-8), illustrate different waveforms at different frequencies comparable with the initial waveform at 1Hz. As exposed, using these equations with higher frequencies has been proven. In this simulation number of cells express each segment and each parameter in the previous model per cell, that lead to the analogy to the transmission line theory.

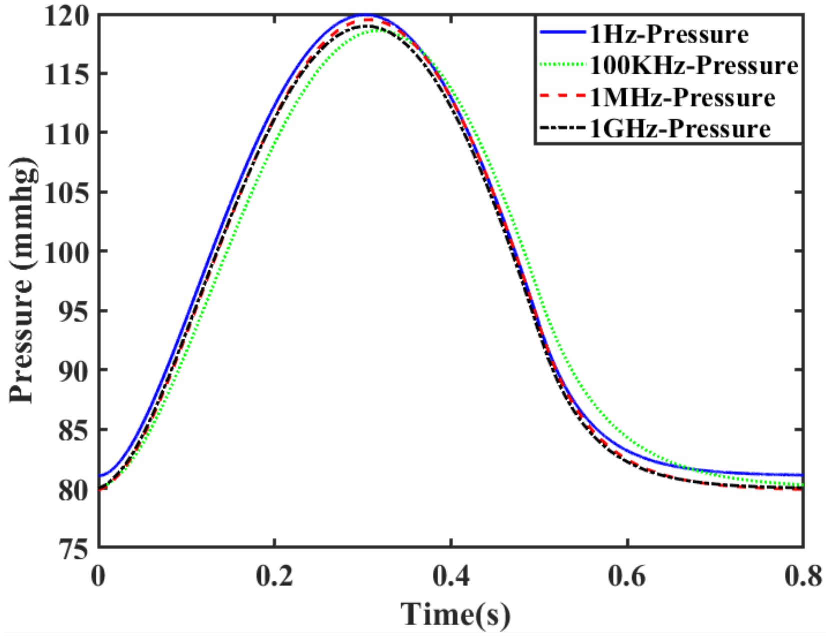


Figure 3-8 Waveform at different frequencies.

3.2 Conclusion

This chapter has been proved the effect of frequency increasing on the lumped elements, and how the same result are achieved depending on certain coherent process. The study of the frequency scaling presents the length effect in this model so the analogy to the transmission line becomes noticeable. In the electric circuits, the voltage is the same at all points on the wire. However, when the voltage changes in a time interval, it takes the signal to travel the wire, so the length becomes important and the wire should be treated as a transmission line. By another expression, when the signal includes frequency components with corresponding wavelengths comparable to the length of the wire, it should follow the length is greater than 1/10 of the wavelength [36]. Because of the effective small length at high frequencies, this study produces a significant role to demonstrate that the circulatory system can be presented as a miniaturized model at high frequency especially at RF frequencies as a transmission line.

The next chapter will provide a parametric study to choose the type of transmission line that can be used to express these results.

Chapter 4 : Transmission line Model

In chapter three, the escalating study lead to the transmission line analogy, in this chapter will examine the type of transmission line.

4.1 Transmission line theory:

At high frequencies, the wavelength is much smaller than the circuit size, resulting phases variation at different locations in the circuit. This fact is the main difference between the circuit theory and transmission line theory. Accordingly, transmission line is a distributed parameter network, where voltages and currents can vary in magnitude and phase over its length, while ordinary circuit analysis deals with lumped elements, where voltage and current do not vary appreciably over the physical dimension of the elements [36].

The transmission line is often schematically represented as a two-wire line since transmission always have at least two conductors as (Figure 4-1.a).

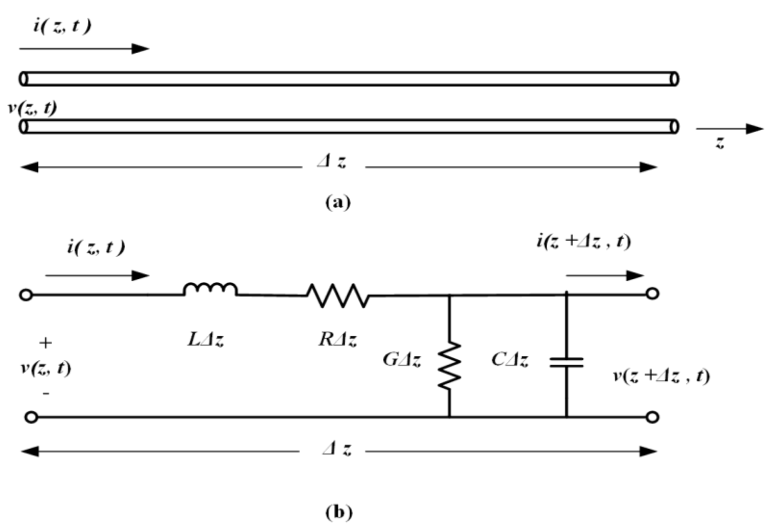


Figure 4-1 Voltage and current definitions and equivalent circuit for an incremental length of transmission line. (a) Voltage and current definitions. (b) Lumped-element equivalent circuit. [36]

The piece of line of infinitesimal length Δz of can be modeled as a lumped-element circuit, as shown in Figure 4-1.b), where R, L, G, and C are per-unit-length quantities defined as follows:

R = series resistance per unit length, for both conductors, in Ω/m.

L = series inductance per unit length, for both conductors, in H/m.

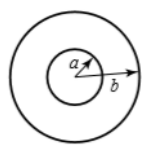
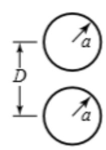
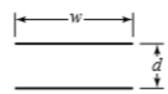
G = shunt conductance per unit length, in S/m.

C = shunt capacitance per unit length, in F/m.

Transmission lines include coaxial cable, a two-wire line, a parallel plate or planar line, a wire above the conducting plane, and a microstrip line. Cross sectional views of these lines consist of two conductors in next table. Each of these lines consists of two conductors in parallel. Coaxial cables are used in electrical laboratories and in connecting T.V sets to T.V antennas, Microstrip lines are important in integrated circuits where metallic strips connecting electronic elements are deposited on dielectric substrates [37].

For the coaxial, two-wire transmission lines, and the parallel plate, the distributed parameters are related to the physical properties and geometrical dimensions as Table 4-1) [36]:

Table 4-1 : Represent the parameters of different transmission line

	COAX	TWO-WIRE	PARALLEL PLAT
			
L	$\frac{\mu}{\pi} \ln \frac{b}{a}$	$\frac{\mu}{\pi} cosh^{-1}(\frac{D}{2a})$	$\frac{\mu d}{w}$
C	$\frac{2\pi\epsilon'}{\ln b/a}$	$\frac{\pi\, \epsilon'}{\cosh^{-1}\ (\frac{D}{2a})}$	$\frac{\epsilon' w}{d}$
R	$\frac{R_S}{2\pi}(\frac{1}{a}+\frac{1}{b})$	$\frac{R_S}{\pi a}$	$\frac{2R_S}{w}$
G	$\frac{2\pi\omega\epsilon''}{\ln b/a}$	$\frac{\pi\, \omega\, \epsilon''}{\cosh^{-1}\ (\frac{D}{2a})}$	$\frac{\omega\epsilon'' w}{d}$

4.2 Microstrip Model

Microstrip line is considered as the most common useable planer transmission line because of the simplicity, size, the fabrication processes low cost, also the flexibility to work with different types of devices.

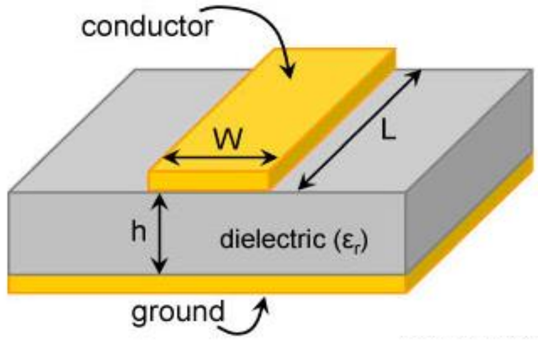


Figure 4-2 Microstrip line illustration.

A microstrip transmission line is shown in the (Figure 4-2). Its physical characteristics include the microstrip width (*w*), the microstrip thickness (*h*), the substrate height (*d*), and the relative permittivity constant (ε_r).

Studying transmission line parameters variant with frequency is important. These parameters are estimated at a particular frequency of interest to avoid errors in design or analysis. Typically, for microstrip line, the frequency variation of the effective dielectric constant is more significant than the variation of characteristic impedance, in terms of both relative change and the relative effect on performance. A change in the effective dielectric constant may have a substantial effect on the phase delay through a long section of line, while a small change in characteristic impedance has the primary effect of introducing a small impedance mismatch. In addition, the complexity of modeling these effects, approximate formulas are generally useful only for a limited range of frequency and line parameters, and numerical computer models are usually more accurate and useful.

Here, very simple equations are presented for the microstrip-line electrical parameters, such as impedances, effective dielectric constants, and attenuation including the losses of the substrate. In microstrip design, it is often observed that the physical circuit performance differs significantly from that theoretically calculated. This is due to factors such as attenuation dispersion, and discontinuity effect that again are functions of physical parameters and the actual circuit layout. The optimal microstrip design should be based directly upon physical dimensions. This requires accurate models and design; the model must be in an easily calculable form. In the proposed design, a Matlab study was used at different (ε_r) range from (2 to 20) and different thickness (0.025 to 8 mm) to define best fitting to the model equations between the relative permittivity and inductance and capacitance were used, to examine the best fitting for our parameters. Two cases in order to define the w/h correction, inductance. Hammerstad provided well-accepted formulas for calculating the effective permittivity and characteristic impedance of microstrip lines [38][39]:

$$A = \frac{Z0}{60}\sqrt{\varepsilon_r + 1/2} + \frac{(\varepsilon_r - 1)}{\varepsilon_r + 1} * (0.23 + \frac{0.11}{\varepsilon_r}) \quad (4.1)$$

$$B = \frac{377\pi}{2Z0\sqrt{\varepsilon_r}} \quad (4.2)$$

The case of the w<2 so,

$$w = \frac{8e^A}{e^{2A}} - 2 \quad (4.3)$$

In case of w > 2 so other formulas will be used,

$$w = \frac{2}{\pi(B - 1 - \log(2B - 1)} + \frac{\varepsilon_r - 1}{2\varepsilon_r \log(B - 1)} + 0.39 - \frac{0.61}{\varepsilon_r} \quad (4.4)$$

After defining, the w/h effective permittivity ε_{eff} can be calculated as:

$$\varepsilon_{eff} = \frac{\varepsilon_r + 1}{2} + \frac{\varepsilon_r - 1}{2} * \frac{1}{\sqrt{(1 + 12 * w/h)}} \quad (4.5)$$

The microstrip L and C considered as the following formulas:

$$L = \frac{Z0}{c0 * \sqrt{\varepsilon_{eff}}} \quad (4.6)$$

$$C = \frac{1}{c0 * Z0 * \sqrt{\varepsilon_{eff}}} \quad (4.7)$$

4.3 Impedance study:

In most transmission lines, the effects due to L and C tend to dominate because of the relatively low series resistance and shunt conductance. Its lossless and equivalent line describes the propagation characteristics of the line. In chapter three the relationship between the lumped element and frequency were presented as:

$$L(f) = \frac{L_1}{f^2} \tag{4.8}$$

$$C(f) = C1 \tag{4.9}$$

In this case of lossless impedance will be:

$$Z = \sqrt{\frac{L1/_{f^2}}{C1}} \tag{4.10}$$

In other words,

$$Z = \sqrt{\frac{L1}{f^2 * C1}} = \sqrt{\frac{\mathrm{L1}/_{f}}{f * C1}} \tag{4.11}$$

We suppose that the defined rules in eq (4.8) and (4.9) are special case of more general formulas.

$$L(f) = L1/(f * \alpha_1) \tag{4.12}$$

$$C(f) = C1 * \alpha_2 \tag{4.13}$$

On this consideration so that which is the maximum value for α_1 is *f* *(equivalent to 4.8)* and minimum value is 1, and the same consideration for C*(f)* assume that it was multiple by $\alpha_2 = 1$ when $\alpha1$ is f , so it will be multiple by f when α_1 is 1. Therefore, there is a flexibility in the impedance, and it can follow many customs but under a condition in the maximum

$$(\alpha_1 * \alpha_2 = f).$$

To maintain this result should have the same β which is $w\sqrt{LC}$ and that can be gained when the conception follows the same condition$(\alpha_1 * \alpha_2 = f)$. The simulated output signal with ADS show the same result in all the combination of LC if the new rules are respected.

To precise this result parametric study was done to validate same form in both Z and β, and in the same time which match the microstrip impedance study. When the previous definitions of characteristic impedance are used in the parametric study, so it will assess the explicit condition for the best characteristic impedance that applied to the equations of the transmission line and predicts parameters values [40]

(Figure 4-3) show that it is possible to control the impedance to practical value by changing the frequency. With frequency greater than 1MHz the impedance is lower than 100 Ohm. (Figure 4-4) show the propagation constant with LC based on equations (4.8) and (4.9). The proposed modification with correction coefficient α1 and α2 can be used to control this issue.

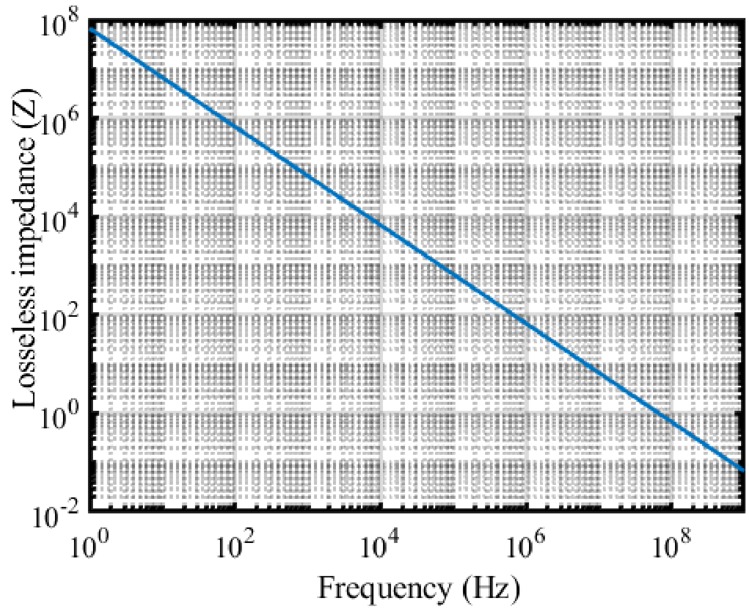


Figure 4-3 Lossless impedance with frequency

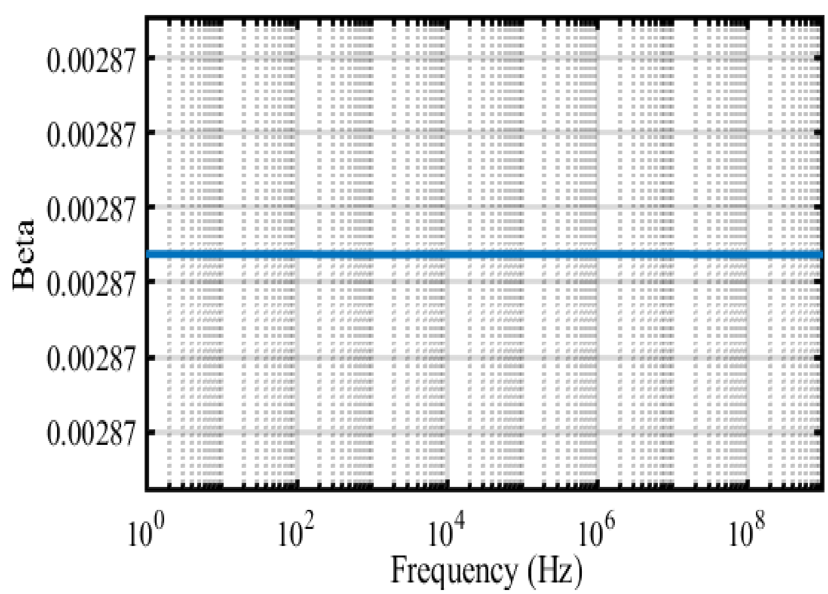


Figure 4-4 Propagation constant with frequency

4.4 Design study:

As shown in chapter three the simulation gives a good result at different frequencies until 1GHz, which known with radio frequencies and define as the frequencies between the upper of audio frequencies and the lower of infrared frequencies, and their range from 20 KHz to around 300 GHz. In this design, 20 MHz frequency is used because it is applicable in term of impedance and can generate using the lab pulse generator, but at the same time, the frequency can be higher as an RF circuit and the parameters can be found.

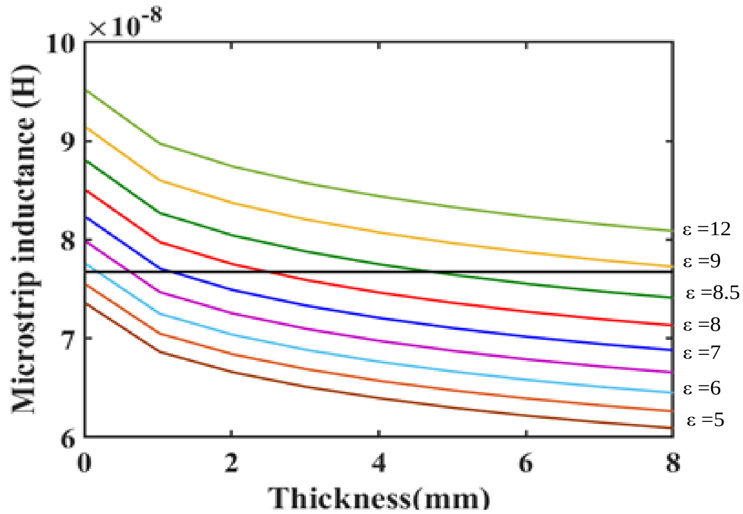


Figure 4-5 Microsrtrip line inductance at different permitivity ε and the model inductance.

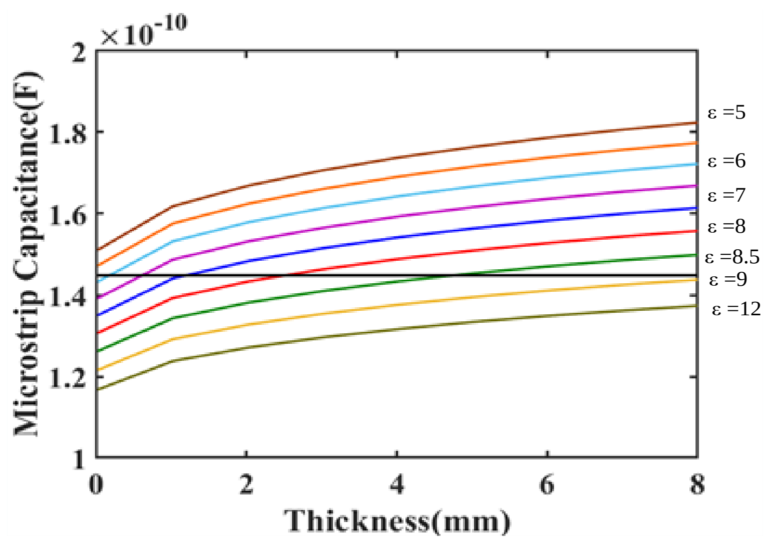


Figure 4-6 Microsrtrip line capacitance at different permitivity ε_r and the model capacitance

The (Figure 4-5) represents the microstrip inductance at different permittivity with the optimal model inductance. The ideal model inductance crosses the studied substrates at different permittivity with different thickness. The same for the capacitance shown in (Figure 4-6). That means there is more than one substrate allowable to be used.

The R parameter should be used to select the thickness. The equivalence Microstrip parallel plate is used to estimate the RG value. The surface resistance is defined by:

$$R_S = \sqrt{\omega\mu/_{2\sigma}} \tag{4.14}$$

(Figure 4-7) shows the thickness of 2.8mm is the proper thickness. The used surface resistivity of the copper walls is (σ = 5.8×107 S/m)

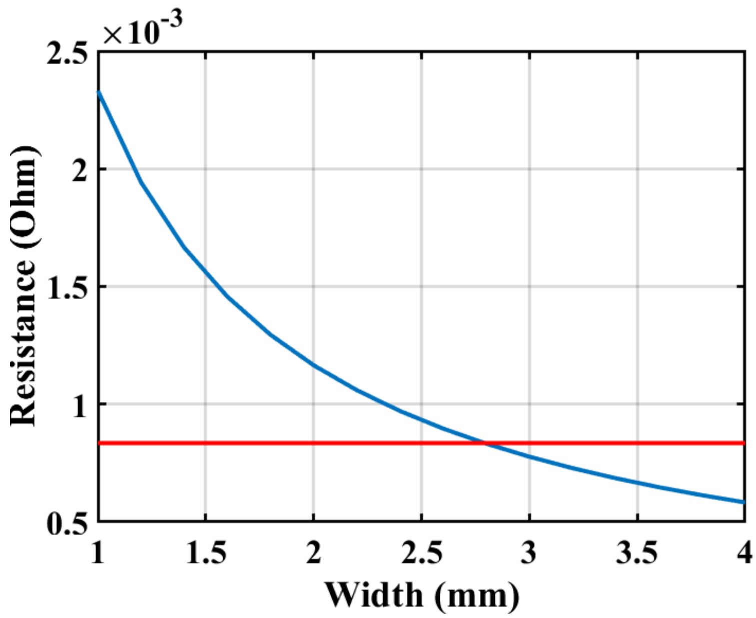


Figure 4-7 Microstrip resistance at different width (Blue) and model resistance(Red).

After defining the required thickness, the width of the strip line should be calculated.

According to (Table 4-1), the parallel plate formula will used, divided by two to have micro-strip formula.

From the equations (4.6), (4.7) and (4.12) the required parameter of this model are

$\boldsymbol{\varepsilon_r}$=8, substrate width equal to 2.8mm, and selected thickness equal to 0.065mm with tangent loss= 0.0024.

4.5 Selected Material Verification

After applying these parameters on ADS simulation, they give an acceptable result, as shown in following (Figure 4-8).

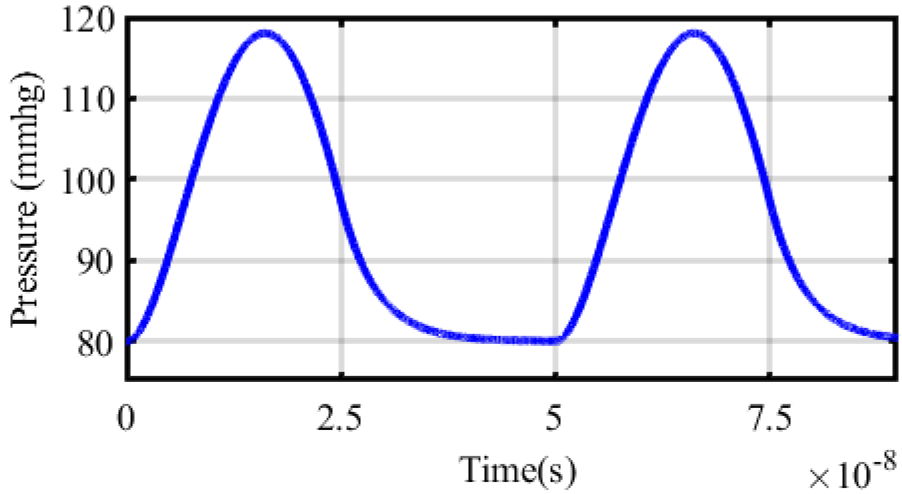


Figure 4-8 Output signal with Microstrip

To validate the variation of the velocity of flow through the blood vessel is it the same with the proposed model, the substrate will be examined at different *cp* and different radiuses. The (Figure 4-9) and (Figure 4-10) illustrate the results to confirm the ability to use the same substrate with different diameters.

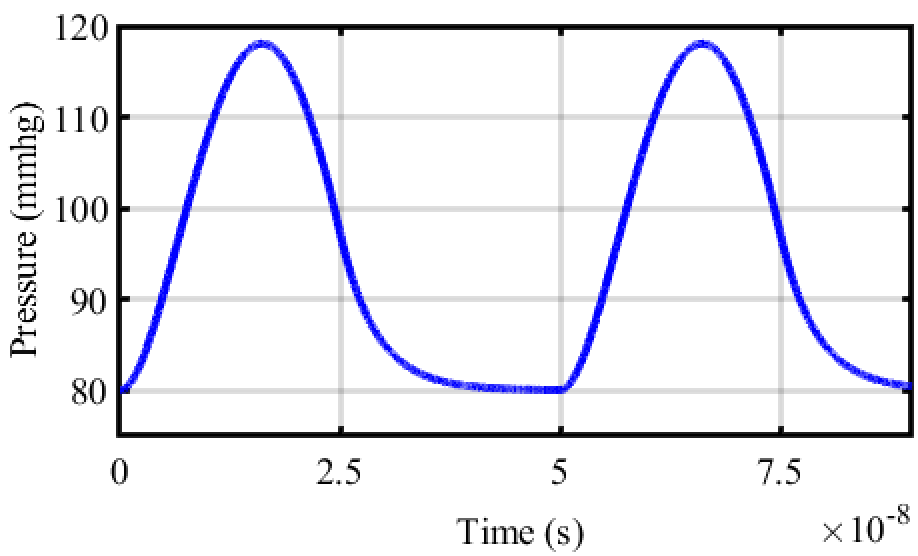


Figure 4-9 Output result with optimized microstrip for *cp*=12 and r=3mm

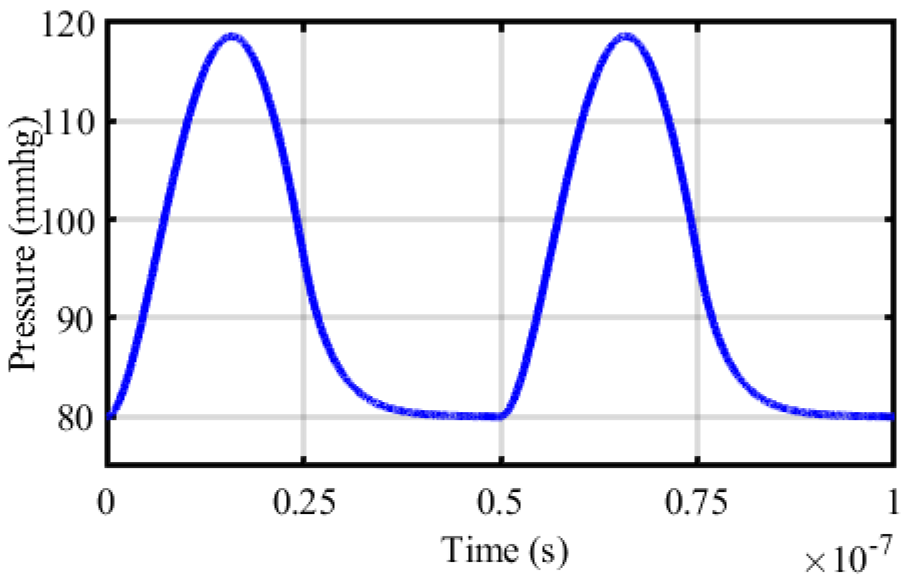


Figure 4-10 Output result with optimized microstrip for *cp*=18 and r=1mm

4.6 Experimental Validation

The fabrication of the proposed microstrip was done using RT/Rogers 6010. To reduce the permittivity air hole are added with estimated diameter periodicity to reach the adequate permittivity. (Figure 4-11) show the fabricated microstrip lines with connected coaxial launcher used for characterization.

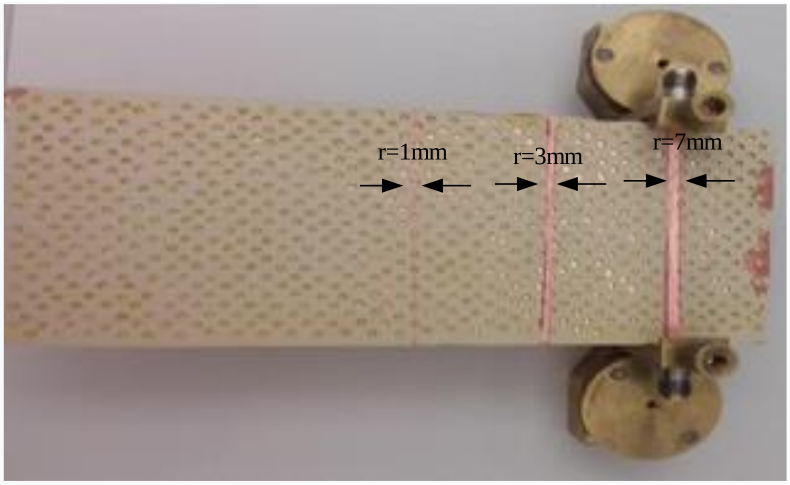


Figure 4-11 Fabricated microstrip line model

The input signal was done using a (81133A) pulse generator at 20 MHz, presented in (Figure 4-12) .The obtained input signal is shown in (Figure 4-13) that close signal to the real input signal. However, the pulse width ratio control is limited.

Figure 4-12 The (81133A) pulse generator

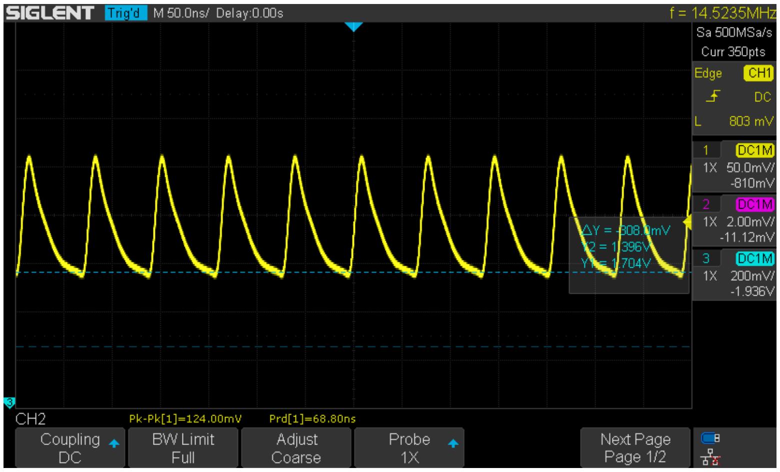


Figure 4-13 The input signal presented on the oscilloscope

(Figure 4-14) and (Figure 4-15) show the result when vessel radius equal 7 mm with corresponding microstrip line width is 2.8 mm as demonstrated in section (4.2) and section (4.4) Even the signal has some noise, but the achieved signal considered as an acceptable result. The input signal cannot be controlled to generate adequate waveform. To overcome this limitation, the microstrip is characterized using VNA (Vector Network Analyser) and the S-parameters box is used in ADS with the sinusoidal modified input signal to estimate the output pressure.

Figure 4-14 Oscilloscope illustrate the result at r=7mm

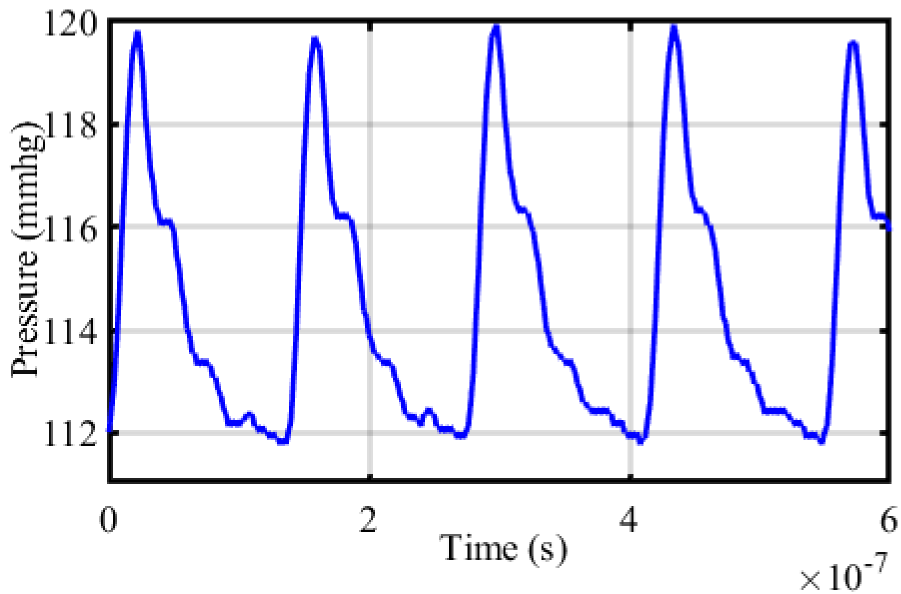


Figure 4-15 The result obtained of microstrip at r=7mm

The model was measured as S-parameter data and the (Figure 4-16) shows the result obtained by Advanced Design System (ADS). The systole and diastole optimal value are reached.

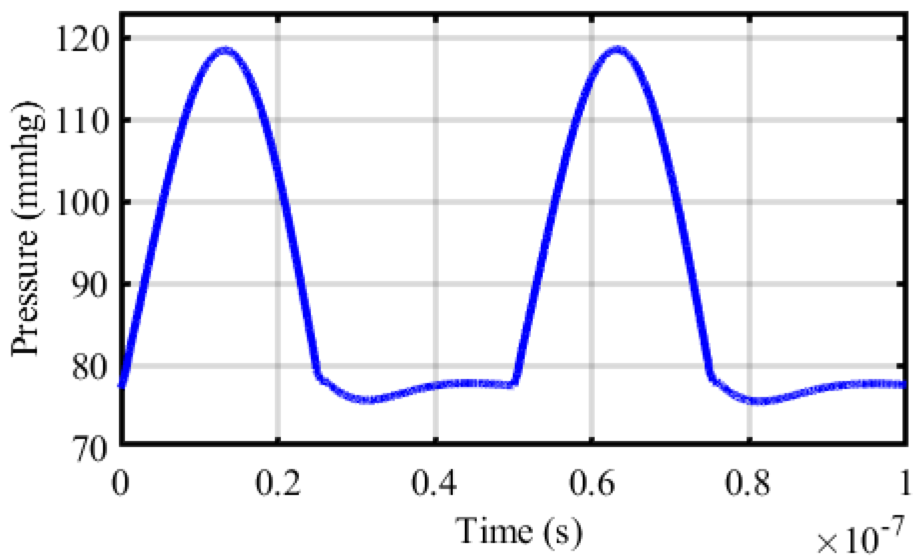


Figure 4-16 Output signal from measured S-parameters result

This chapter has been offered some of the design calculations and how it can work with simulation and theoretical model, parametric study has been done to defining the substrate coefficients. Applying these parameters on ADS schemes and it work perfectly. To certify the proposed model, it was examined with different vessel diameters and wave velocities. The microstrip has been fabricated after the measurement.

4.7 Microstrip Model potential

The prior figures represent the effectiveness of the substrate at different dimensions of vessels. Moreover, if the vessels are connected as (Figure 4-17) by simulation as shown in Figure 4-18), satisfied result in term of shape and values. Pressure drop is notified, and this is what exactly happened in the human body, in the small vessels, the pressure is drops and the flow is increased.

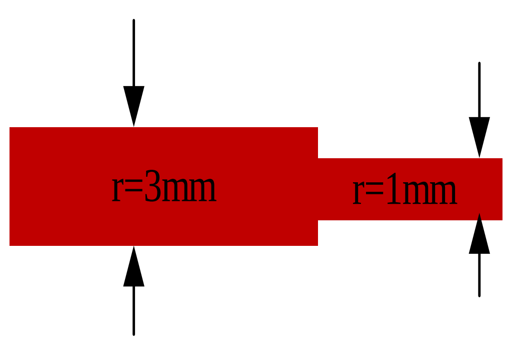


Figure 4-17 Vessels with different radius are connected

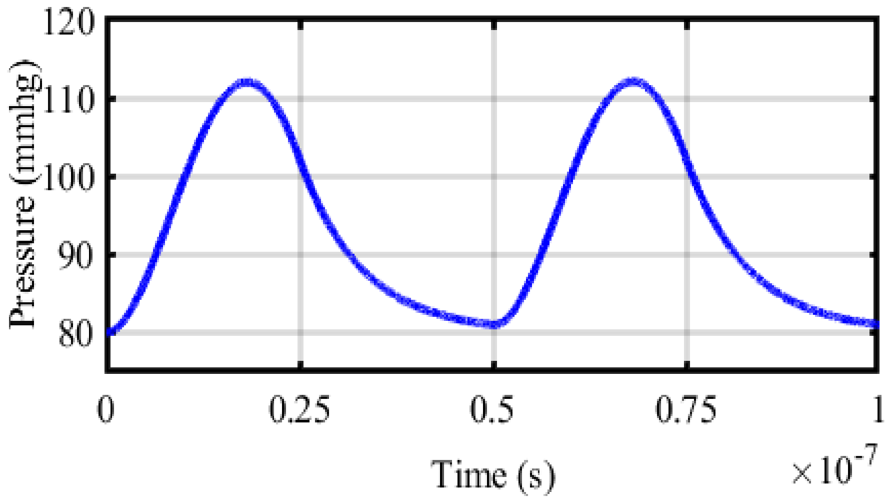


Figure 4-18 Different diameters vessel pressure

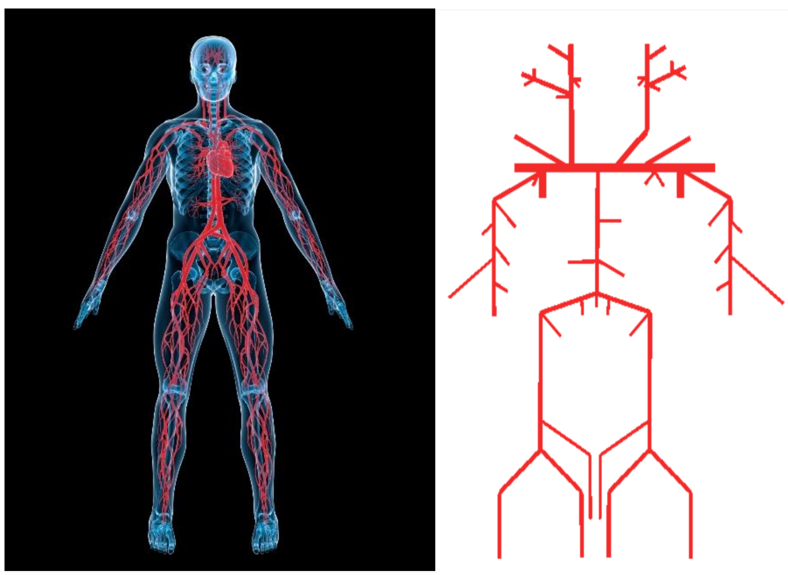


Figure 4-19 The entire circulatory system using transmission line layout

Depending on these results, the entire circulatory system as the illustrated in (Figure 4-19) can be easily modeled with microstrip model, simulated with ADS and validate by measurement. To build model based on lumped element model or mechanical one, complicate and large model is required as the shown in (Figure 1-11).

Conclusion

Circulatory system illnesses are one of the serious global problems especially in the developed countries. The full understanding of the blood flow and blood pressure in the arterial system is influenced by many factors, hence there are numerous models of the arterial blood flow that have been extensively studied in recent years, the unique system for each patient impede to define an actual data for individuals. Thus, the problem transferred from physical environment to the computationally easily handled one. Modeling the human system or subsystem perform an important role to recognize the influence of each system and evaluate possible outcomes for individuals.

Modeling the cardiovascular system is one of main tools for medical innovations, by establishing a better configuration allows studying the mechanisms of blood circulation to define body status under physiological conditions. As well, to evaluates the effect of pathologic changes in the organism and possible impact of therapy without any invasion.

Several methods have been developed to simulate the circulatory system depending on the analogy between the hemodynamic system and the electrical system. Although there are models were created with several different optimized configurations, but most of the simulation had not been accurate and sometimes depending on approximations.

Lumped element based are one of the most popular models. Their result differs from one to another, because some models focused on fitting the output curve without demonstrate the effect of the changing in the dimensions or the blood features. The other models, depending on fluid equations, provide a reasonable result. However, lumped parameter models, are useful to study the relationship of cardiac output to peripheral loads, but because of the finite number of lumped elements, they cannot model the higher resolution aspects of the system without adding many more elements. Consequently, starting a study to demonstrate the hypothesis model, by subjecting the study of transmission line model based on the equivalence between Naiver-stokes equation and Maxwell equations.

The study was started by simulating an initial circuit at 1Hz which equivalent to the heart frequency by ADS. Next, the frequency was increased until 100Hz and the effect of the frequency scaling have been revealed on the lumped elements and some formulas have been established as mentioned in chapter three. These formulas have been proven at higher frequency, and they were presented an acceptable result. This study composed a significant role to demonstrate model that the circulatory system can be presented as a miniaturized model at high frequency especially at

RF frequencies as a transmission line. Due to the effective small length at high frequencies this lead to the proposed hypothesis. Subsequently, the type of transmission line chooses as microstrip, and examined with different material by a parametric study at different permittivity/thickness at practical frequency of 20 MHz This study showed the microstrip parameters that were closer to the model theoretical parameters. This provides a flexibility to the design, due to the different thickness and width. After the parametric study was done, the substrate was defined with εr equal to 8 and thickens equal to 0.065mm.

The fabrication of the proposed microstrip was done using RT/Rogers 6010. Air holes were needed to reduce the permittivity, so it has been added with estimated diameter and periodicity to reach the sufficient permittivity. The lab measurement has been determined with pulse generator shown in (Figure 4-12), the result was acceptable excluding the noise influence slightly affect the waveform result; however, the S-parameter showed more satisfied. Thus, the hypothesis has been proven and the model has been exposed, and that lead to a practical model with low cost and small size.

This model will be a great tool for understanding the mechanisms of blood circulation, so it is the initial stage to demonstrate the whole system. Therefore, the future work is designing the entire human body using the transmission line depending on clinical data. Firstly, the passive part includes the whole circulatory system and nervous system, and the active part that include the heart and liver and the other organs. In addition, changing in the pressure waveform can be one sign of a defining some health problem as shown in chapter three the effect of each lumped element indicates to a specific issue. Experimentally, after using the tuning, we notice that the changing on R1 is affected on systole cycle at the same time the diastole cycle influenced by L, here the issue can be signed to the blood characteristics such as (blood clotting); also, Y and C can represent the alerting in the vessel size. Forward and backward traveling pulses also can be predicted. From this information, the diseases can be diagnosed after applying the transmission line theory on the model in accordance with a complete complex design supported by the intelligent artificial network. The achieved microstrip model, provide an acceptable result, a more accurate model will be observed, once a real measured input will be provided with a real clinical data. The ability to estimate the pressure waves accurately and noninvasively would indeed be a valuable addition to current non-invasive tools such as those provided by echo/Doppler and magnetic resonance imaging.

References

[1] Agency of Canada, public health. (2009). Facts about heart disease. Retrieved from http://www.hricanada.org/about-heart-disease/facts-about-heart-disease

[2] Abecia, S. L. (n.d.). Biology 2201 Unit 3 – Maintaining Dynamic Equilibrium I Notes Human Circulatory System the Human Circulatory System. Retrieved from https://www.academia.edu/10688338/Biology_2201.

[3] Guyton, Arthur C., and John E. Hall. "Medical physiology." (1961).

[4] Martini, F. H., Nath, J. L.,and Bartholomew, E. F. (2011). Fundamentals of Anatomy and Physiology. Benjamin Cummings, ninth edition)

[5] McGraw-Hill Reason. Biology11. chapter12 The Circulatory System (pp. 476–520).

[6] Heart tee four chamber view.jpg. (2018, August 11). Wikimedia Commons, the free media repository. Retrieved 17:15, September 24, 2019.

[7] Hall, J. E. (2015). Guyton and Hall textbook of medical physiology e-Book. Elsevier Health Sciences.

[8] High blood pressure (hypertension). (n.d.). Retrieved from https://www.bupa.co.uk/health-information/heart-blood-circulation/high-blood-pressure-hypertension.

[9] Frank, Otto,(1899) ``Die Grundform des arteriellen Pulses'', Zeitung für Biologie 37 pp.(483-586).

[10] Kokalari, I., Karaja, T., and Guerrisi, M. (2013). Review on lumped parameter method for modeling the blood flow in systemic arteries. Journal of biomedical science and engineering, 6(01), 92.

[11] De Pater, L., and Van den Berg, J. W. (1964). An electrical analogue of the entire human circulatory system. Medical electronics and biological engineering, 2(2), pp(161-166).

[12] De Pater, L. (1966). An electrical analogue of the human circulatory system (Doctoral dissertation, Bronder-Offset).

[13] John B. W.,(2008) Ibn al-Nafis, the pulmonary circulation and the Islamic Golden Age. Journal of Applied Physiology. 105(6), 1877-1880.

[14] Ribatti, D. (2009). William Harvey and the discovery of the circulation of the blood. Journal of angiogenesis research, 1(1), 3.

[15] Bernoulli, D. (1738). Hydrodynamica: sive de viribus et motibus fluidorum commentarii. Johannis Reinholdi Dulseckeri.

[16] Euler, L.,(1775) Principles for the movement of blood through the arteries dealings. Posthumous works by detecting mathematics and physics in 1844, 2, pp(1814-1823).

[17] Parker, K. H. (2009). A brief history of arterial wave mechanics. Medical & biological engineering & computing, 47(2), pp.(111-118).

[18] Noordergraaf, A., Verdouw, P.D. and Boom, H.B.K.,(1963) The use of an analog computer in a circulation model. Progress in Cardiovascular Diseases, 5, 419439.

[19] Westerhof, N., Bosman, F., De Vries, C.J. and Noordergraaf, A.(1969), Analog studies of the human systemic arterial tree. Journal of Biomechanics, 2, 121143.

[20] Avolio, A. P.(1980), Multi-branched model of the human arterial system. Medical and Biological Engineering and Computing, vol(18), pp.(709-718).

[21] Zamanian, S. A. (2007). Modeling and simulating human cardiovascular response to acceleration (Doctoral dissertation, Massachusetts Institute of Technology).

[22] Watrous, R. L. (2009) A computational model of cardiovascular physiology and heart sound generation. Conference Proceedings IEEE Engineering in Medicine and Biology Society, 3105-3110.

[23] Debbaut, C., De Wilde, D., Casteleyn, C., Cornillie, P., Van Loo, D., Van Hoorebeke, L., ... and Segers, P. (2012). Modeling the impact of partial hepatectomy on the hepatic hemodynamics using a rat model. IEEE Transactions on Biomedical Engineering, 59(12), PP.3293-3303.

[24] Hoppensteadt, F. C., and Peskin, C. S. (2012). Modeling and simulation in medicine and the life sciences (Vol. 10). Springer Science & Business Media.

[25] P Segers, E R Rietzschel, M L De Buyzere, N Stergiopulos, N Westerhof, L M Van Bortel, T Gillebert and P R VerdonckThree- and four-element Windkessel models: Assessment of their fitting performance in a large cohort of healthy middle-aged individuals.

[26] Lambermont, B., Gerard, P., Detry, O., Kolh, P,Potty, P., Defraigne, J. O., D’Orio, V., and Marcelle, R.(1997)Comparison between three- and four algorithm yielded a WK4-p model that showed a element Windkessel models to characterize vascular behaviour that was very similar to that of the WK3 properties of pulmonary circulation. Arch. Physiol. Biochem., vol(150),pp.(625–632)

[27] Creigen, V., Ferracina, L., Hlod, A., van Mourik, S., Sjauw, K., Rottschäfer, V., Zegeling, P. (2007). Modeling a heart pump. European Study Group Mathematics with Industry, 7.

[28] Yoshigi, M., Knott, G. D., and Keller, B. B. (2000). Lumped parameter estimation for the embryonic chick vascular system: a time-domain approach using MLAB. Computer methods and programs in biomedicine, 63(1), pp.(29-41).

[29] Çinar, Y., Şenyol, A. M., & Duman, K. (2001). Blood viscosity and blood pressure: role of temperature and hyperglycemia. American journal of hypertension, 14(5), pp (433-438).

[30] Kenner T. (1989). The measurement of blood density and its meaning. Basic Res Cardiol. Mar-Apr; 84(2):111-24

[31] Ghasemalizadeh, O., Mirzaee, M. R., Firoozabadi, B., and Hassani, K. (2014). Exact modeling of cardiovascular system using lumped method. arXiv preprint arXiv:1411.5337.

[32] M Mirzaee, O Ghasemalizadeh, B Firoozabadi.(2008) Modeling Vessels in Human Cardiovascular System and Calculating Pressure Wastes Using Lumped Method. The Internet Journal of Bioengineering. vol 3 no 2.

[33] C. Debbaut et al.(2011), "From Vascular Corrosion Cast to Electrical Analog Model for the Study of Human Liver Hemodynamics and Perfusion," in IEEE Transactions on Biomedical Engineering, vol. 58, no. 1, pp. 25-35.

[34] Catanho, M., Sinha, M., and Vijayan, V. (2012). Model of aortic blood flow using the Windkessel effect. University of California of San Diago, San Diago.

[35] Taylor, M. G. (1959). An experimental determination of the propagation of fluid oscillations in a tube with a visco-elastic wall; together with an analysis of the characteristics required in an electrical analogue. Physics in Medicine & Biology, 4(1), 63.

[36] Pozar, D. M. (2009). Microwave engineering. John Wiley & Sons.

[37] Edwards, T., and Steer, M. (2000). Foundations of interconnect & microstrip design. MICROWAVE ENG EUR, (pp.51-52).

[38] Hammerstad, E., and Jensen, O. (1980, May). Accurate models for microstrip computer-aided design. IEEE MTT-S International Microwave Symposium Digest (pp. 407-409).

[39] Awasthi, Y. K., Singh, H., Sharma, M., Kumari, S., and Verma, A. K. (2017). Computer-aided design-based circuit model of microstrip line for terahertz interconnects technology. The Journal of Engineering,vol (9), (pp.512-526).

[40] Rautio, J. (1991.). A new definition of characteristic impedance. IEEE MTT-S International Microwave Symposium Digest. doi: 10.1109/mwsym.147116