Mechanical effects of cochlear implants on residual hearing*

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Abstract: Cochlear implant is a surgically implanted electronic device that can provide a sense of sound to a person who is severely hard of hearing. The effect of a cochlear implant on residual, low frequency, hearing is complex and not well understood. This research focuses on changes of the cochlear mechanics due to a cochlear implant by comparing the basilar membrane, BM, response before and after the implantation using a computational model of the cochlea. In the model, cochlea implants were introduced into the lower cochlear fluid chamber and the active amplification process of the cochlear is not considered, since a passive cochlear model whose response does not depend on stimulus level can reasonably well represent the cochlea for subjects with hearing impairment. The results for the basilar membrane velocity show that the volume change in the fluid chambers due to the implant has a little effect, less than 3 dB at low frequencies, on the basilar membrane velocity. A more extreme condition, in which the cochlear implant is assumed to touch the basilar membrane at some or the whole positions and thus impeded its motion, was also studied. Although there is no travelling wave propagating in the basal region in the latter case, the remainder of the cochlea is still coupled to the stapes by the incompressible fluid. The basilar membrane velocity at low frequencies is relatively unaffected by the blocking of the basilar membrane motion in the basal region, although the effect is more dramatic for excitation frequency whose characteristic place is close to the end of the implant. Although this work does not model every aspect of the hearing loss after cochlear implantation as measured clinically, it does provide a way of predicting the possible mechanical effects of the implantation on the cochlear passive mechanics and residual hearing.

Keywords: Cochlear implant; residual hearing; mechanical effect; cochlear model

人工耳蜗术后力学特性改变对残余听力影响分析

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摘 要:人工耳蜗是一种植入式电子设备,是目前为重度、极度听力损失患者重建听力的最行之有效的临床治疗方法。人工耳蜗术后患者残余听力的水平取决于多种复杂因素,尚未被完全了解。本研究聚焦在人工耳蜗植入后对于耳蜗动力学特性产生的影响。通过建立的生物力学模型对比人工耳蜗植入前后基底膜响应,进而得到可能的听力损失情况。所采用的耳蜗模型为被动式,不存在对低声压级信号的放大过程,这是由于大部分人工耳蜗患者的听力阈值较高,耳蜗在高声压级信号激励时其响应具有线性、无主动增益特性。计算结果显示,由于人工耳蜗植入带来的淋巴液体积变化对基底膜振动速度影响不大,在低频范围内最多造成 3 dB 听力损失。还假设了两种极端情况,即部分或者全部植入的电极与基底膜相接触并阻止其产生响应。在全部电极与基底膜接触情况下,虽然接触部分的基底膜无法运动,但是镫骨位置的激励还是可以通过耳蜗内的不可压缩流体传递至基底膜剩余完好位置。在这些极端假设情况下,低频激励下的基底膜响应并未受到太大影响,仅是电极末端位置对应的听

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觉特征频率受影响较大。虽然研究不能解释造成人工耳蜗植入后残余听力损失的全部原因,但研究结果表明人工耳蜗植入带来的耳蜗动力学特性改变对残余听力的影响不大。

关键词: 人工耳蜗;残余听力;力学效应;耳蜗模型

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1　Introduction

There are about 360 million people worldwide [1] who have disabling hearing loss and an estimated 30 million of those are in China. There were no effective treatments for profound hearing loss until the advent of cochlear implant (CI), which is a surgically implanted electronic device that can provide a sense of sound to a person who is severely hard of hearing.

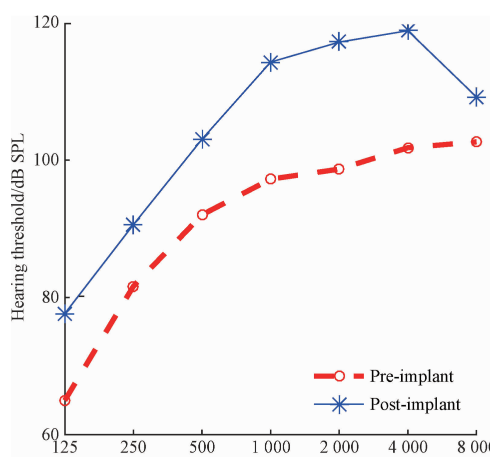
Over 180 000 people worldwide have received a cochlear implant, with approximately 40,000 recipients (70% are children) in China. Cochlear implants already give considerable benefit in terms of speech understanding to severely and profoundly hearing impaired adults and children[2-3] and recent work has combined the use of cochlear implants at high frequencies with residual hearing at low frequencies to significantly increase the number of patients who could benefit from such devices. Several problems remain with cochlear implants, whose design is generic and not tailored to the individual. The next stage of development will require personalized devices, whose geometry and excitation strategy is targeted to the individual patient. A major problem with cochlear implants is the broad variability in outcomes, in which even patients using exactly the same implant system can have very different scores in tests. The geometry of the cochlea varies considerably between patients [4-6]. This increases complexity in choosing cochlear implants and difficulty in surgery, and greatly increases risk of damaging patients' residual hearing [7-10].

Studies show that the overall effect of cochlear implant insertion on a patient's residual hearing is an increase of the minimum detectable level by 25 dB for low frequencies and less than 5 dB for high frequencies. The reason for this reduction of residual hearing is complex and dependent on many factors such as operation method, patient's condition etc. With an increasing number of cochlear implant patients, a model that is able to predict the possible effects on the hearing level before and after surgery will be particularly useful to clinicians [11-14]. Cochlear implant candidates by definition have very little measurable hearing, thus can be assumed to have lost cochlear active amplification function.

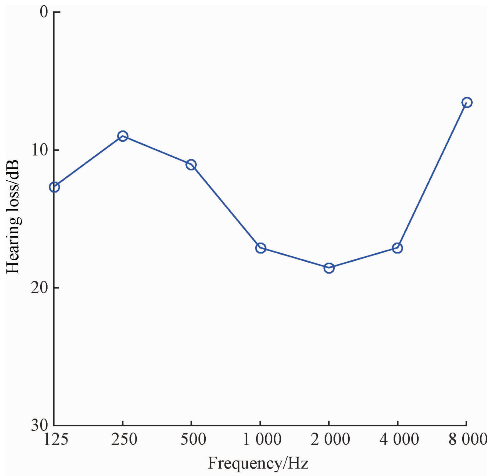
In this paper we present preliminary work using a computationally efficient model of the cochlea to predict the possible mechanical effects due to cochlear implants on residual hearing and the basilar membrane response. The important aspect of this in the prevention of hearing loss is that not only is there a concern about the implantation interfering with residual hearing in patients, there is also a long-term concern to preserve the cochlear fine structures after implantation in patients who are currently profoundly deaf. This is in the hope that regenerative or gene treatments will become available at some time in the future [15].

2　Model of the human cochlea

Fig. 1 (a) shows average hearing threshold measured from 24 cochlear implant users before and after surgery (on average 6.75 months post-implantation) when the cochlear implant is not turned on. The overall effect of a cochlear implant is an increase of the hearing loss value about less than 10 dB for low- and high-frequencies to about 20 dB for most mid-frequency region, as shown in Fig. 1 (b). This residual hearing reduction is due to complex reasons or a combination of multiple factors. The volume change of the cochlear fluid and mechanical interference to the basilar membrane (BM), due to a cochlear implant, could play a role in the reduction of residual hearing after implantation. A cochlear model, which will be used to study possible mechanical effects of the cochlear implant, will be described in details in this section.



(a) Hearing threshold



(b) Hearing loss

Fig. 1 Average value of hearing threshold and hearing loss (Data were collected from cochlear implant recipients at Beijing Children's Hospital, Capital Medical University)

The model target is the human cochlea with high hearing thresholds, so it is not a bad approximation to use a passive linear model of the cochlea to study possible influences due to existence of a cochlear implant. A passive model means the nonlinear and active processes due to the hair cells are not included. Most descriptions of the mechanical response of the cochlea depend on the forward propagation of a single "slow" wave[16-17], due to an interaction between the inertia of the fluid in the chambers of the cochlea and the stiffness of the basilar membrane, and can be reproduced using box models with simplified geometry[18]. Previous study [11] shows that the coupled linear mechanical response of the cochlea can be represented by a model with only a single dimension along the cochlea length by predefining a radially profile for the basilar membrane velocity. This linear cochlear dynamics can be divided into two components: 1) the pressure distribution determined by the fluid coupling within the cochlear chambers when driven by the basilar membrane and stapes velocities, and 2) the basilar membrane dynamics respond to the imposed pressure distribution.

Although the described model may over simplify the real cochlea, it is, however, able to reproduce the basic cochlear travelling wave. The basilar membrane dynamics in this model are represented by a series of isolated mass-spring-damper systems, as shown in Fig. 2, whose resonance frequency is tuned to match the frequency-position map of the human cochlea. The natural frequency of each system is tuned to vary from 20 000 Hz at the base down to 20 Hz at the apex.

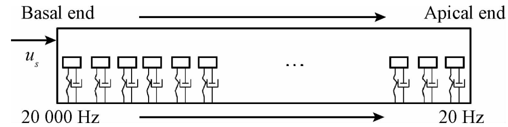


Fig. 2 Idealized representation of the cochlea representing the basilar membrane as a series of mass-spring-damper systems distributed along the cochlear length

In order to actually represent the basilar membrane dynamics, the single longitudinal variables for modal pressure difference and modal velocity need to be spatially sampled as finely as required, for example at least six elements within the shortest wavelength present [19]. By dividing the cochlear model into N segments, we can define, at a given frequency, the complex pressure and velocity vectors, $\boldsymbol{p}$ and $\boldsymbol{v}$, along different cochlear locations to be:

$$\boldsymbol{p} = [p(1), p(2), \cdots, p(N)]^{\mathrm{T}} \tag{1}$$

$$\boldsymbol{v} = [v(1), v(2), \cdots, v(N)]^{\mathrm{T}} \tag{2}$$

The basilar membrane, however, is assumed only to extend from element 2 to element $N-1$. The first element of $\boldsymbol{v}$, $v(1)$, is the normalized stapes velocity, defined as the stapes volume velocity divided by the elemental area, ΔW, where Δ is the length of an element in the x direction, which is equal to L/N where L is the overall length of the cochlea. The final element, N, is used to account for the behavior of the helicotrema, which equalizes the pressure in the two chambers at the end of the cochlea, so that $p(N)$ is 0. The vector of pressures due to the vector of stapes and basilar membrane velocities can be written as:

$$\boldsymbol{p} = \boldsymbol{Z}_{\mathrm{FC}}\boldsymbol{v} \tag{3}$$

where $\boldsymbol{Z}_{\mathrm{FC}}$ is a matrix of impedances due to the fluid coupling. The vector of basilar membrane velocities can also be written as:

$$\boldsymbol{v} = \boldsymbol{v}_{\mathrm{s}} - \boldsymbol{Y}_{\mathrm{BM}}\boldsymbol{p} \tag{4}$$

where $\boldsymbol{v}_{\mathrm{s}}$ is a vector whose first element is equal to the normalized stapes velocity, u_{s}, unloaded by the pressure in the cochlea, with all other elements being 0. $\boldsymbol{Y}_{\mathrm{BM}}$ is a matrix of basilar membrane admittances.

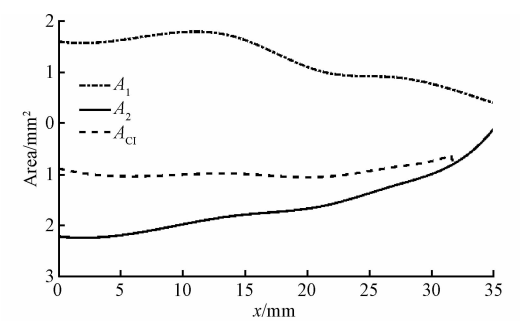
Equations (3) and (4) can be combined to give a simple expression for the vector of basilar membrane velocities in the coupled cochlea as:

$$\boldsymbol{v} = [\boldsymbol{I} + \boldsymbol{Y}_{\mathrm{BM}}\boldsymbol{Z}_{\mathrm{FC}}]^{-1}\boldsymbol{v}_{\mathrm{s}} \tag{5}$$

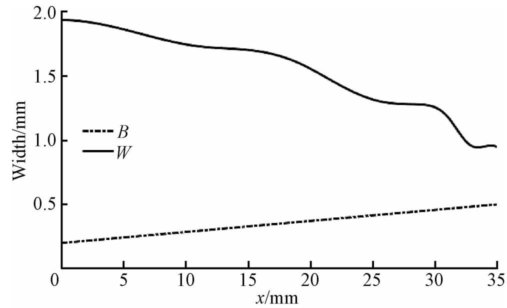
where $\boldsymbol{v}_{\mathrm{s}}$ is the known input driving the stapes. Thus, once the form of $\boldsymbol{Y}_{\mathrm{BM}}$ and $\boldsymbol{Z}_{\mathrm{FC}}$ have been determined, the coupled response for a given stapes velocity can be readily calculated using simple linear algebra.

3　Implementation of a cochlear implant

In the real cochlea, the cross-sectional area varies with position and should be taken into account in the model. Fig. 3 shows variations of cross-sectional area of the two fluid chambers, scala vestibuli (SV), A_1, scala tympani (ST), A_2, and modified scala tympani, A_{CI}, when a cochlear implant is included, along the length of the human cochlea, together with corresponding assumed variations in the width of the fluid chamber, W, and basilar membrane width, B. In this model, the scala media, SM, is assumed to be merged into the SV, since the Reissner's membrane that separates the SM from SV is often assumed to be "acoustically transparent" having no influence to the cochlear mechanical functions[20]. The human cochlea variations are based on data given by Zakis J et al[21], which are interpolated using a cubic spline function and are reasonably consistent with the measurements of Thorne M et al[22] and the earlier estimates provided by Fletcher H[23] and Zwislocki J J[24]. The cochlear implant used for modifying the scala tympani area was chosen to be the longest one clinically used in order to maximize the influence of fluid volume change due to implantation.



(a) Variation of the cross-sectional area along the human cochlea, together with a modified area, A_{CI}, when a cochlear implant is inserted



(b) Variation of the cochlear partition width and basilar membrane width

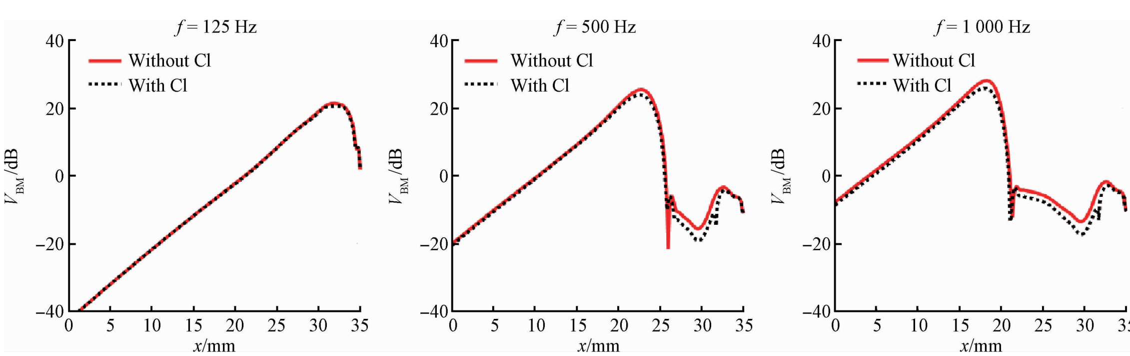
Fig. 3　Variations of the human cochlea geometry

Besides volume change due to the cochlear implant, the dynamics of the basilar membrane could also be altered. To implement this type of effect, we assumed the basilar membrane admittance, Y_{BM}, is 0 over the locations that are being occupied by the cochlear implant.

4　Mechanical effect of the cochlear implant on residual hearing

4.1　Change of fluid volume due to the cochlear implant

Fig. 4 shows the calculated distribution of the coupled basilar membrane velocity using the model described above. In this condition, a cochlear implant is introduced into the lower chamber, having a length of 31.5 mm with a tapering cross-sectional area varies from 1.3 mm^2 at the base down to 0.2 mm^2 at the apex. The results for the coupled response, with and without the effect of implant on the cochlear fluid volume, differ by less than 3 dB at low frequencies, indicating that this volume change has a negligible effect on the passive behavior of the modeled cochlea at frequencies below 1 000 Hz, as shown in Fig. 5.



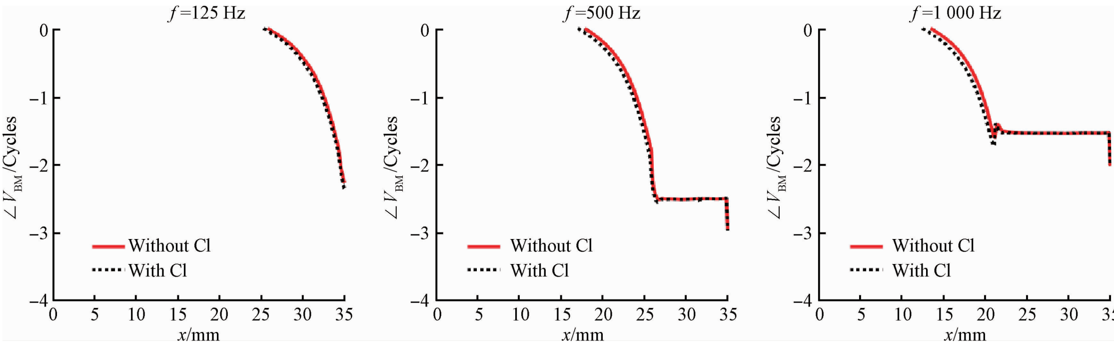


Fig. 4 Coupled basilar membrane velocity distribution in the model at different frequencies when the volume of the fluid chamber is assumed to be changed due to the cochlear implant

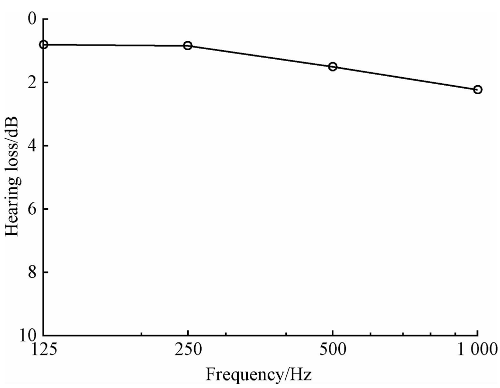


Fig. 5 Hearing loss due to fluid volume change caused by insertion of a cochlear implant shown in Fig. 3

4.2 Change of basilar membrane due to the cochlear implant

Fig. 6 shows the coupled basilar membrane velocity when part of the basilar membrane is assumed to be blocked, which is 5 mm long in this example. It can be seen that at low frequencies when the blocked part is away from the characteristic place, the coupled basilar membrane velocity is not significantly affected by the cochlear implant, except the blocked region. The hearing losses due to this partially blocking are between 1 ~ 3 dB, as shown in Fig. 7, which indicates that the partially blocking does not contribute to the overall hearing loss, as shown in Fig. 1, significantly.

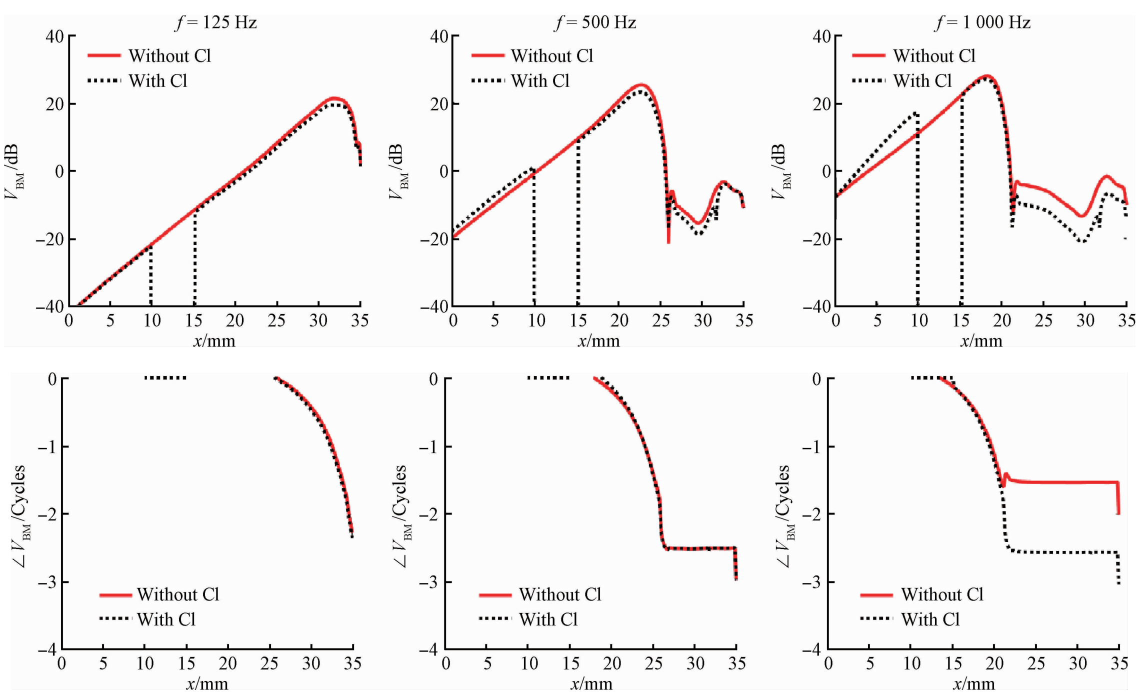


Fig. 6 Coupled basilar membrane velocity distribution in the model at different frequencies when the basilar membrane motion is assumed to be partially blocked due to the cochlear implant

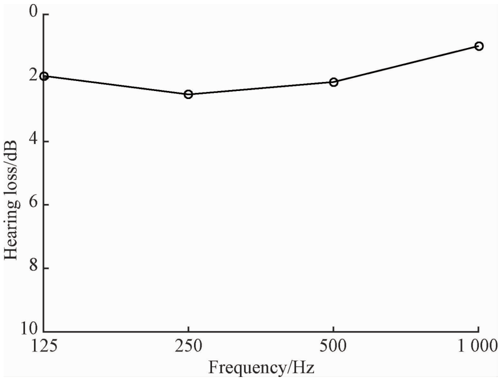


Fig. 7　Hearing loss due to the basilar membrane is blocked by part of the cochlear implant

4.3　Blocking the basilar membrane

A more extreme effect would be expected if the cochlear implant touched entirely the basilar membrane and thus blocked its motion. In order to explore the consequences of this effect, another condition has been simulated in which the cochlear implant is assumed to completely block the motion of the basilar membrane all the way along its 19 mm length. This was modelled by setting the basilar membrane admittance to be 0 for these corresponding basilar membrane elements. In this extreme condition, although the cochlear travelling wave cannot now propagate in the basal region, as shown in Fig. 8, the remainder of the cochlea is still coupled to the stapes by the incompressible fluid. The travelling wave now starts from 19 mm along the cochlea, as seen in the phase responses. The hearing loss caused by this entire blocking becomes more significant than the partially blocked case, as seen in Fig. 9. For these low frequency simulations the passive cochlear model, however, may not be fully representative since the measured average hearing thresholds are less than 80 dB, indicating the cochlear active amplification may still be functional. This suggests an active cochlear model would be desired for these low frequency regions.

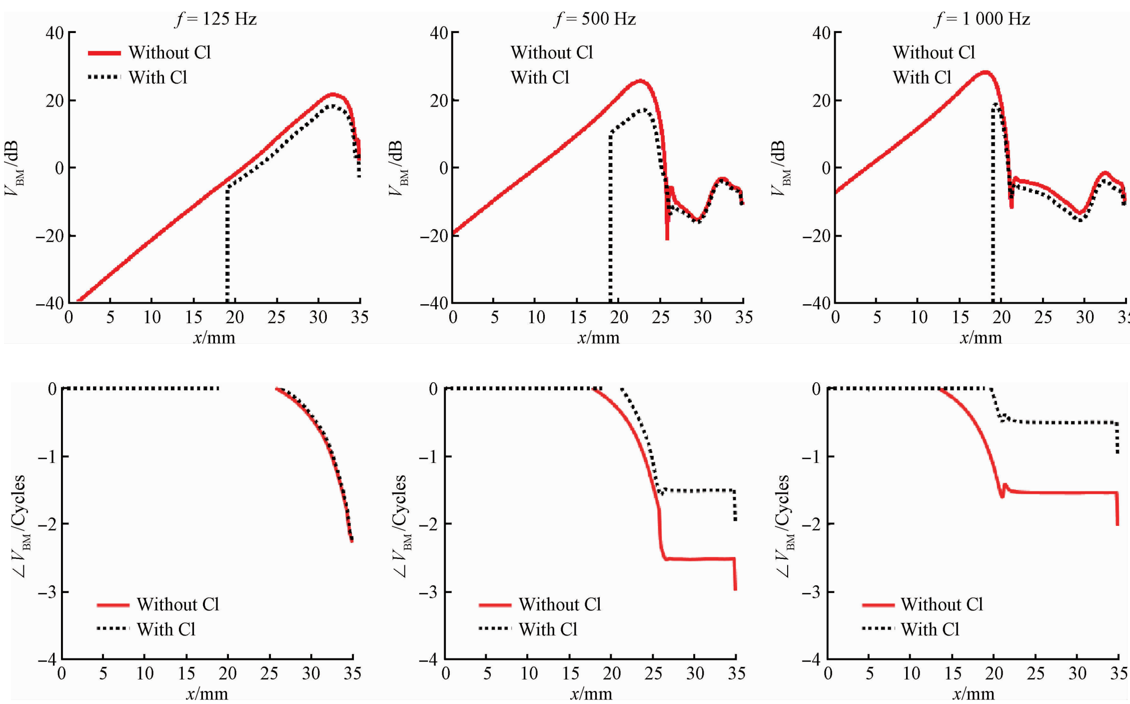


Fig. 8　Coupled basilar membrane velocity distribution in the model at different frequencies when the basilar membrane motion is assumed to be blocked along the entire length of the cochlear implant

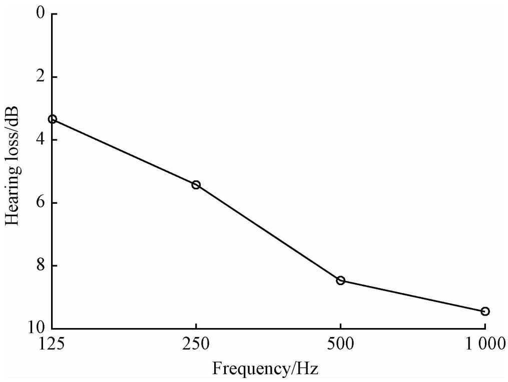


Fig. 9 Hearing loss due to the basilar membrane is blocked over the entire length of the cochlear implant

5 Conclusion and discussion

The effect of a cochlear implant on residual hearing is an important clinical topic, but is not yet fully understood. This preliminary work predicts some possible mechanical effects due to a cochlear implant on the basilar membrane velocity, thus the causes of loss in residual hearing, using a simple passive cochlear model.

The fluid volume change due to the implant leads to about 3 dB hearing loss at low frequencies suggesting that volume change plays a negligible role in affecting the passive basilar membrane response.

If the basilar membrane was partially blocked by the cochlear implant, the effect of this mechanical interference on the peak of the basilar membrane velocity is predicted to be small at low frequencies. The change of the basilar membrane dynamics due to the interface of the implant, especially when the basilar membrane is entirely blocked by the implant, is predicted to dramatically affect the cochlear response at frequencies higher than the characteristic frequency that corresponds to the characteristic place close to and beyond the end of the implant.

Although this passive model of the cochlea does not represent every aspect of the human cochlea with implantation, it does provide a way of predicting the possible mechanical effects of the implantation on the cochlear passive mechanics and the residual hearing. This work cannot explain 10 ~ 20 dB hearing loss as shown in Fig. 1, but it clearly shows that there must be other, presumably physiological, reasons that cause this damage to the residual hearing.

In the future, the cochlear active amplification will be introduced into the model and may provide a better representation of the residual hearing and furthermore a better prediction of the mechanical effect of a cochlear implant at low frequencies.

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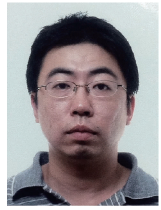

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