KEMP ECHO DIGITAL FILTERS

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Abstract:

This paper reviews the digital filters associated with Kemp Echoes, where the latter are outer ear reflections resulting from incident pulses applied to the outer ear but characterizing the inner ear. It is thought that the Kemp Echoes can be used for noninvasive testing and diagnostics of the inner ear.

I. Introduction

In 1978 Kemp [1, abstract] announced that "a new auditory phenomenon has been identified in the acoustic impulse response of the human ear." This response has been checked by others [2] and, consequently, has become to be known as the "Kemp Echo." An example is shown in Fig. 1 where the horizontal axis is scaled to sampling units (about 100 samples per ms) and the vertical represents pressure (the initial peak being about 100 Nm^{-2} and is indicative of reflection from the middle ear [for about 100 samples following the first peak is what one might get from a test cavity the remainder being primarily from the inner ear]). Although of small amplitude the Kemp echo can be isolated with good filtering techniques. In normal cases the Kemp echo lasts for milliseconds but in some cases, as appears to be the case for tinnitus, can last for very much longer. Because there is a significant difference in the Kemp echo for healthy versus certain classes of damaged ears, it is felt that the Kemp echo can be used to give a noninvasive means to quickly and easily characterize some types of damage to the inner ear [3]. Consequently, it is appropriate to develop models of the ear that can be used to simulate Kemp echoes in their impulse response and from which a characterization of the inner ear can be made. Here we discuss a digital filter simulation technique relevant to Kemp echo theory.

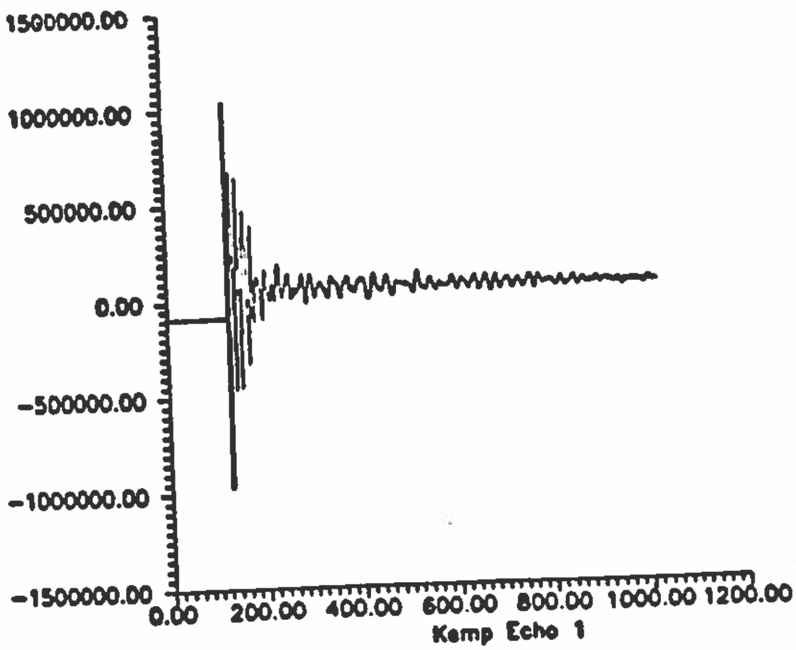

Figure 1 A Kemp Echo. Courtesy of Dr. H.P. Wit, Institute of Audiology, Groningen

Although many models of the inner ear exist [4] most of these are not directly appropriate to the Kemp echo phenomena since the latter is presently based upon fluid wave motion in the cochlea. Hence a treatment in terms of scattering vari- ables is the most relevant, and it is through the use of such that we proceed. By way of notation, z denotes the z-trans- form variable with systems variables expressed in terms of z though this dependence is often omitted to avoid cluttering the equations. By way of structure of the ear and standard modeling of the ear we refer to [4] and [5]. In our treat- ment the inner ear is assumed to vary in one direction, x, and this too is discretized at values x_k, which will usually be represented solely by the integer variable k (with again the variable often left off for simplification). As depicted in Fig. 2, the cochlea is assumed to be made of two equal halves, the vestibular one (denoted in the following by sub- script v) and the tympanic one (denoted by subscript t), each of cross section S(x) and separated by a basilar membrane which has a vibrating part of transverse length D(x) (which increases as the basilar membrane narrows in going away from the end nearest the ear drum). At the other end of the coch- lea, away from the ear drum, the two halves of the cochlea are joined by an opening, the helicotrema, which allows fluid to flow between the two halves. For the human to hear there is also signal flow between the the ear and the brain, for which we assume that the neural output is proportional to the pressure difference along the moving portion of the basilar membrane.

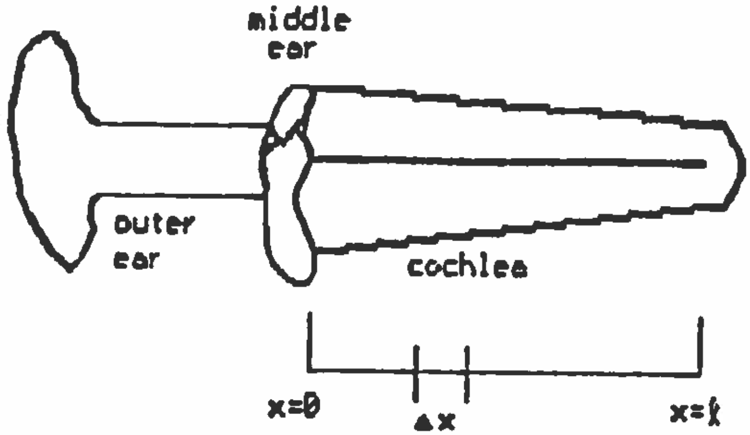

Figure 2 Unfolded Idealized One-dimensional Form of Cochlea

II. A Digital Scattering Model

Considering only the mechanical motions, the digital (synthesis filter) structure we choose to represent the inner ear is a cascade of M sections of identical structure but differing parameters loaded in a direct connection as shown in Fig. 3 [6][7] and fed by a source. The load represents the helicotrema and the source represents the excitation to the

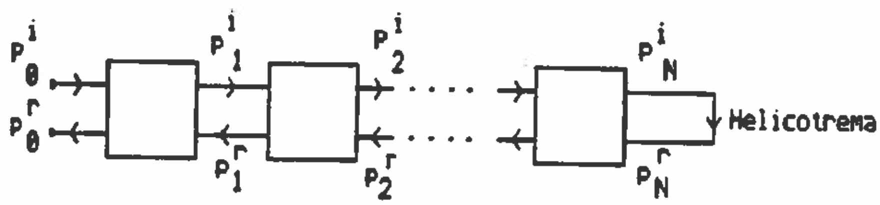

Figure 3 Cascade Digital Filter for Use with Kemp Echos

outer ear as transferred through the transformer action of the (middle ear) ear drum system to be seen by the inner ear. One set of variables that can be used to represent these sections is the pressure difference across the basilar membrane, $p=p_v-p_t$ and the area velocity $u_v=Sv_v$, where v_v is the cochlea fluid velocity, which, by conservation of mass and incompressibility of the fluid, is equal to $-v_t$ (since we are assuming for simplicity that $S_v=S_t$). However, for a scattering-type treatment we choose to work with incident-type and reflected-type pressures, p^i and p^r, respectively. These quantities are taken to be related via

$$\begin{bmatrix} p \\ u_v \end{bmatrix} = \begin{bmatrix} 1 & 1 \\ G & -G \end{bmatrix} \begin{bmatrix} p^i \\ p^r \end{bmatrix} \qquad (1)$$

where G = G(x,z) is the (complex-valued) "conductance" normalization parameter which we assume to be held constant (in x) within a section.

Each of the non-load (signal-flow graph) sections in Fig. 3 represents a discretized volume element of the inner ear of short length δx as represented in Fig. 4 (for simplicity of exposition we assume all δx equal). The kth section shown in Fig. 4 can be described by its transfer scattering matrix using the incident and reflected pressure variables (indexed with k to denote space discretization and as functions of z to denote time discretization). Thus, it has the inverse transfer scattering matrix

$$\begin{bmatrix} p^i \\ p^r \end{bmatrix}_k = \frac{1}{\tau_k}\begin{bmatrix} 1 & -\rho_k \\ -\rho_k & 1 \end{bmatrix}\begin{bmatrix} e^{-\gamma_k} & 0 \\ 0 & e^{+\gamma_k} \end{bmatrix}\begin{bmatrix} p^i \\ p^r \end{bmatrix}_{k-1} \qquad (2a)$$

where

$$\tau_k = 1 - \rho_k \qquad (2b)$$

and the "reflection coefficient" ρ_k and the "propagation constant" γ_k are given in terms of fluid mechancial properties of the kth volume element [8]. In (2a) the first (right-hand) square matrix factor represents an attenuated delay for the section while the second factor represents the mechanical motion of the basilar membrane as it affects the fluid flow. As this is simply a model, we could actually reverse the order of these two matrix factors and essentially achieve the same results.

As derived from the continuous time fluid mechanical model of the cochlea [6][8], we have

$$\rho(x,z) = d[\ln(Z_s Z_p)]/4dx \qquad (3a)$$
$$\rho_k(z) = \rho(x_k,z)\cdot\delta x \qquad (3b)$$
$$\gamma(x,z) = (Z_s/Z_p)^{1/2} \qquad (3c)$$
$$\gamma_k(z) = \gamma(x_k,z)\cdot\delta x \qquad (3d)$$

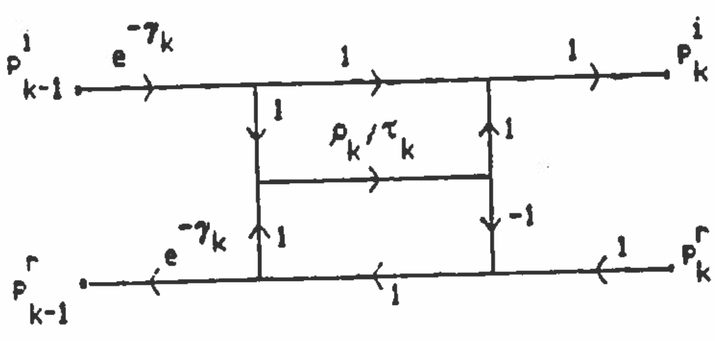

Figure 4 Digital Filter Section

where

$$Z_s(x,z) = 2[R+mf(z)]/S(x) \tag{3e}$$

$$Z_p(x,z) = [\mu f(z)^2+\sigma(x)f(z)+\phi(x)]/[D(x)f(z)] \tag{3f}$$

in which, for the fluid, R is the coefficient of frictional force and m is its mass density, while for the basilar membrane μ is the mass, σ is the friction, and ϕ is the spring constant, all three being per unit area. f(z) is the sampling function used for converting from continuous time to discrete time. Physically Z_s represents a series-arm impedance to fluid flow as we travel (longitudinally) down the line while Z_p represents a parallel-arm shunting impedance due to the basilar membrane motion. We have also that $G^2=1/(Z_s Z_p)$, i. e. G is the characteristic admittance at the point x (we take the positive branch of the square root for z within the unit circle so that a passive normalization is used). Since f(z) represents differentiation in the time domain we could use backward differencing to give (with δt the uniform difference in sampling times)

$$f(z) = (1 - z^{-1})/\delta t \tag{3g}$$

though other functions, such as the bilinear transformation, are possibly more useful [9, p. 113].

Many measurements have been made on the human ear to determine the parameters of the cochlea, but there is still not total agreement in the literature. We take the following functional dependencies upon x for a healthy ear [10][11, p. 27-30]

$$D(x) = D_0 + D_1 x \tag{4a}$$
$$S(x) = S_0 \exp(S_1 x) \tag{4b}$$
$$m = \text{constant},\ R = \text{constant},\ \mu = \text{constant} \tag{4c,d,e}$$
$$\sigma(x) = \sigma_0 \exp(-\sigma_1 x) \tag{4f}$$
$$\phi(x) = \phi_0 \exp(-\phi_1 x) \tag{4g}$$

The problem one would like to solve via the Kemp echoes is: What are these 0 & 1 subscripted constants for a given healthy ear, or, for a given ear that has damage, how do the x dependencies deviate from what they should be for an equivalent healthy ear?

Summarizing the model, we take M sections of the type of Fig. 4 in cascade to represent the scattering behavior of the human inner ear with parameters assumed of the form presented above. Except for the initial peaks, the Kemp echo is assumed to be the reflected signal resulting from the impulse as an incident excitation of this synthesis filter.

III. Determination of the Model Parameters

Given a Kemp Echo, or better a set of equivalent ones that can be averaged, we wish to determine the parameters of the digital filter mentioned above. For this we use an analysis digital filter derived from the synthesis filter of Fig.

3 by turning around the reflected signals (i.e. by reversing the arrows on the bottom input and output lines of Fig. 4). In this case (2a) becomes directly the scattering matrix of the analysis section and the quantities on the right are known for k=0. Estimation theory is then used to determine the parameters of the section and the section terminal variables for k=1, for which the process is repeated for k=2, ..., M. One determines M by requiring p^i_M to be close to p^r_M according to some error criterion; if truly the model is a good one, such will result. In order to find the parameters of the sections themselves, one can use PARCOR theory [9], at least for the reflection coefficient factor of the inverse transfer scattering matrix. This necessitates a splitting into the two matrix factors of (2a) in which case we rewrite (2a) as

$$p^i_k{}' = [\exp(-\gamma_k)]p^i_{k-1} \quad (5a)$$
$$p^r_k{}' = [\exp(+\gamma_k)]p^r_{k-1} \quad (5b)$$

$$p^i_k{}'' = p^i_k{}' + \rho_k \cdot p^r_{k-1}{}' \quad (5c)$$
$$p^r_k{}'' = \rho_k \cdot p^i_k{}' + p^r_{k-1}{}' \quad (5d)$$

$$p^i_k = p^i_k{}''/(1-\rho_k) \quad (5e)$$
$$p^r_k = p^r_k{}''/(1-\rho_k) \quad (5f)$$

The philosophy is to do the estimation upon (5). This because (5c&d) are in the standard lattice form such that normal PARCOR estimation can be used upon them. Once ρ_k and the incident and reflected variables through (5d) are estimated for the kth section from the (k-1)st data, then (5e&f) are used to obtain the incident and reflected variables for the next, (k+1)st, section. In doing this one needs a method for estimating γ_k with the help of (5a&b). Since γ_k is assumed rational it is helpful to take logarithms so that again an estimation rational in the parameters is undertaken. It then is convenient to do a least squares estimate on the logarithms of $p^i_k{}'$ & $p^r_k{}'$.

At present these steps are being undertaken and preliminary results appear to be quite encouraging [12].

IV. Neural Excitation

It is known that excitation of the brain is via motion of the fine hairs attached to the basilar membrane [4, p. 1333], this excitation being transmitted via the auditory nerve [13]. This motion is most certainly excited by the pressure wave that moves through the cochlea, and, hence can be expressed in terms of the change in pressure

$$\delta p_k = (p^i_k + p^r_k) - (p^i_{k-1} + p^r_{k-1}) \quad (6)$$

in going through section k of the digital sections. Consequently, by tapping the signals at the section terminals, as shown in the modification of Fig. 4 given by Fig. 5(a), we can monitor neural-type excitations [14]. Of further interest is the fact that there is known to be "active feedback" associated with acoustical emissions from the ear [15]. Although this normally is small and the biological source for this is not definitely known, it can be represented by placing further inputs at the section terminals with these inputs controlled by signals via feedback processing, perhaps from the brain, of the neural-type outputs just mentioned, and as illustrated also in Fig. 5(b).

V. Discussion

Since almost all humans lose some hearing with aging due to deterioration of the inner ear, noninvasive determination of the properties of the inner ear should be very important to most of us. The Kemp echo appears to be an ideal tool to accomplish these diagnoses of various hearing diseases and hearing losses. However, the signals are relatively weak, and, hence, rather sophisticated methods must be employed to properly model and interpret clinical results. Overall this appears to be an area where modern systems theory has much to offer. To indicate this we have outlined some of the main ideas and results which we believe will be significant to the development of Kemp echo theory toward a practical clinical tool. This involves the modeling of the ear via cascade digital signal processors for which we have only mentioned here

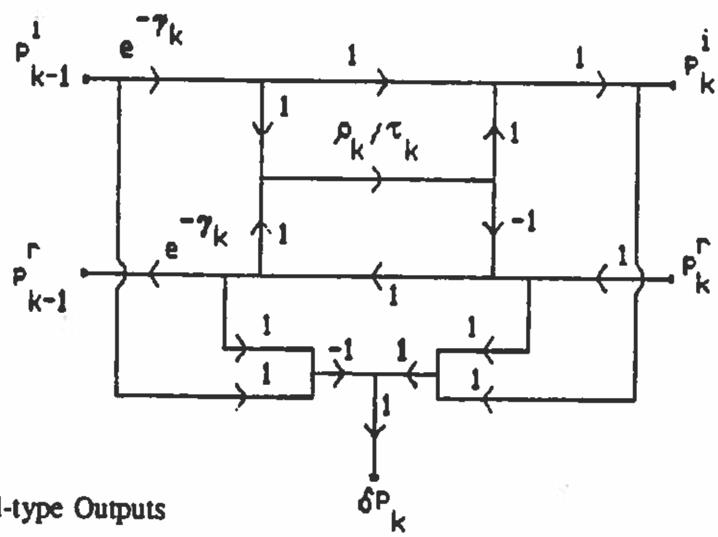

Figure 5(a) Neural-type Outputs

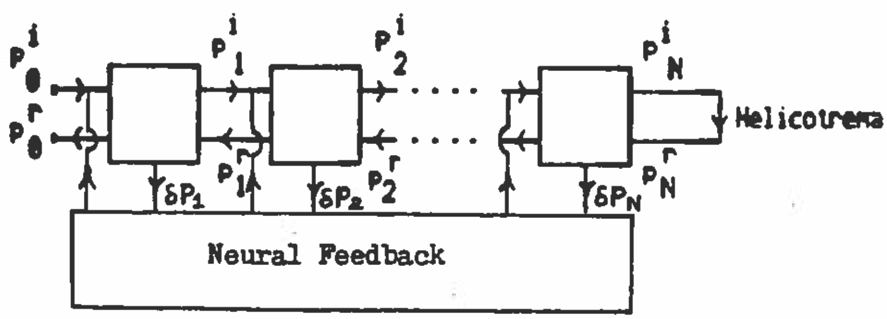

Figure 5(b) Neural Feedback

the linear theory. Since the ear is known to have nonlinear responses and a connection to the brain, it remains to develop adequate nonlinear and neural-type representations, both of which are presently under development.

Here we have omitted the details of derivation which can, however, be found in [16] which also covers the nonlinear case. It should be noted that the derivation of (2) is via a direct discretization of the partial differential equations representing the analog behavior of the scattering-type variables of the cochlea. If one substitutes the results found by the use of (2) back into (1) to recover the pressure and velocity on the right and the left of a junction of sections, one will find discontinuities at the junctions due to the discretization of G and the equality used for the scattering-type pressure variables used on the right and the left of the junction of two sections. This discontinuity can be compensated by incorporating another factor in (2a) that is the product of the transformation coefficient matrix in (1) for one side of the junction times the inverse of the coefficient matrix for the other side of the junction. However, for a large enough number of sections this discontinuity is negligible since G decreases slowly with x (roughly in exponential form). Since x is discretized at the left side of a section, the least error due to discretization should appear by using variables to the right of a junction, which is the left of a section.

It is to be commented that the variables are of scattering type, rather than true scattering variables, since G varies with z. Consequently, one can not apply some of the classical criteria to the resulting quantities. For example, although the mechanical media is passive the resulting scattering matrix need not have the bounded-real characteristics of a passive scattering matrix. This could be remedied by a more complicated choice for (1), but would result in the loss of (3a) and the resulting (2a), which are key to the PARCOR parameter identification techniques being used via (5). Finally we comment that the treatment here has been primarily for the inner ear. A section for the middle ear is primarily a transformer that goes in cascade with those for the inner ear. It can, thus, be absorbed in the driving source, once it is identified [16].

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