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Research paper

Place pitch versus electrode location in a realistic computational model of the implanted human cochlea

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A B S T R A C T

Place pitch was investigated in a computational model of the implanted human cochlea containing nerve fibres with realistic trajectories that take the variable distance between the organ of Corti and spiral ganglion into account. The model was further updated from previous studies by including fluid compartments in the modiolus and updating the electrical conductivity values of (temporal) bone and the modiolus, based on clinical data. Four different cochlear geometries are used, modelled with both lateral and perimodiolar implants, and their neural excitation patterns were examined for nerve fibres modelled with and without peripheral processes. Additionally, equations were derived from the model geometries that describe Greenwood's frequency map as a function of cochlear angle at the basilar membrane as well as at the spiral ganglion. The main findings are: (I) in the first (basal) turn of the cochlea, cochlear implant induced pitch can be predicted fairly well using the Greenwood function. (II) Beyond the first turn this pitch becomes increasingly unpredictable, greatly dependent on stimulus level, state of the cochlear neurons and the electrode's distance from the modiolus. (III) After the first turn cochlear implant induced pitch decreases as stimulus level increases, but the pitch does not reach values expected from direct spiral ganglion stimulation unless the peripheral processes are missing. (IV) Electrode contacts near the end of the spiral ganglion or deeper elicit very unpredictable pitch, with broad frequency ranges that strongly overlap with those of neighbouring contacts. (V) The characteristic place pitch for stimulation at either the organ of Corti or the spiral ganglion can be described as a function of cochlear angle by the equations presented in this paper.

1. Introduction

Modern cochlear implants (CIs) are multi-channel devices that make use of the cochlea's tonotopical organization by having multiple electrodes stimulate different subpopulations of the auditory nerve fibres in the inner ear. The pitch percept resulting from stimulating such a subpopulation depends on the electrode's location in the cochlea. In practice it is presumed that the frequency alignment of the contacts follows the Greenwood function (Greenwood, 1990), which assigns a characteristic frequency to an auditory nerve fibre based on the position of its peripheral tip along the basilar membrane (BM).

There is evidence that CI patients perform better when the electrically stimulated frequency range corresponds to the acoustic input frequency range, and that patients with bilateral implants or combined electrical and acoustical hearing will benefit from having the frequencies of the two devices or modalities correctly matched to each other (Dorman et al., 1997; Shannon et al., 1998; Fu and Shannon, 1999; Baskent and Shannon, 2004, 2005, 2007; Siciliano et al., 2010; Li and Fu, 2010; Yoon et al., 2011, 2013). As a consequence, it becomes important to correctly estimate the pitch elicited by the CI. With electrical stimulation, neurons will often be stimulated directly in the central axon, rather than at the BM. Therefore, due to non-radial trajectories of auditory nerve fibres in the apex (Ariyasu et al., 1989; Kawano et al., 1996; Stakhovskaya et al., 2007) one can expect that the acoustically derived Greenwood function will not suffice to estimate the correct frequency alignment for electrical stimulation.

There have been several studies of pitch matching performed with CI patients with (contralateral) residual hearing, with varying results. Some pitch matching studies showed no systematic

Abbreviations: CI, cochlear implant; eCAP, electrically evoked compound action potential; EFI, electrical field imaging; MPR, multiplanar reconstruction; BM, basilar membrane; OC, organ of Corti; SG, spiral ganglion; XP, excitation profile

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deviations from the Greenwood function (Vermeire et al., 2008; McDermott et al., 2009; Carlyon et al., 2010a), while most studies found pitch matches lower than expected from the Greenwood function (Blamey et al., 1996; Boëx et al., 2006; Baumann and Nobbe, 2006; Reiss et al., 2007; Dorman et al., 2007; Baumann et al., 2011; Schatzer et al., 2013).

Other studies have investigated pitch perception as a function of stimulus level, again with inconsistent results. Shannon (1983) found that in the subject tested for level effects, an increase in stimulation level lead to an increase in perceived pitch. Townshend et al. (1987) tested three patients and reported that for one subject pitch increased with level, while for the other two subjects the pitch decreased. Pulse rate matching experiments done by Pijl (1997) suggested that lower stimulus levels led to higher pitch. Arnoldner presented two pitch scaling studies, the first one (Arnoldner et al., 2006) showed a clear increase in pitch at higher current levels in 10 patients whereas only one patient showed the opposite effect. In the second study (Arnoldner et al., 2008), a pitch increase was found in only 4 patients while for 10 patients pitch decreased with higher stimulus levels. Finally Carlyon et al. (2010b) did a series of experiments to study the effect of pulse rate on pitch perception and concluded that in 16 out of 21 cases, their results were consistent with pitch increasing with signal level, with the remaining 5 cases showing the opposite effect.

The main objective of the present paper is to investigate CI-induced pitch percepts using a computational model of the implanted human cochlea. Although pitch percepts are in reality determined by a combination of place of excitation and temporal cues, this modelling study will focus exclusively on the effects of place pitch, therefore ‘pitch’ will be used as shorthand for ‘place pitch’ throughout the paper.

The computational model used in this study is an updated version of the one used in previous studies by Frijns and co-workers, and employs a realistic three-dimensional spiral tapered geometry of the human cochlea, which is based on histological data (Frijns et al., 2000, 2001, 2009a,b, 2011; Briaire and Frijns, 2000a,b, 2005, 2006; Snel-Bongers et al., 2013). In the studies done before 2009, this realistic cochlear geometry contained radially defined auditory nerve fibres; each fibre ran from the BM/organ of Corti (OC) directly to its cell body in the spiral ganglion (SG), located inside Rosenthal's canal. In other words, the fibres did not traverse apically or basally along the cochlear scalae. However, anatomical data show that in reality, auditory fibres in the apical part of the cochlea follow an oblique course; the fibres there do not traverse from the BM to the modiolus directly, but proceed basally along the cochlear scalae before reaching Rosenthal's canal. In fact, Rosenthal's canal ends at around 1.75 cochlear turns, while the OC is usually about one turn longer (Ariyasu et al., 1989; Kawano et al., 1996; Stakhovskaya et al., 2007), meaning that the cell body of an auditory nerve fibre in the apex can be located as much as a full turn more basally into the cochlea than its peripheral tip. The consequence of this is that if electrodes in the apex stimulate neurons near their cell bodies rather than in the peripheral processes (which is likely, since many of today's implants actually target the SG), they will excite a different fibre population than one would expect based on radial nerve fibre trajectories. This could lead to dramatically different pitch percepts than estimations using the Greenwood function would suggest, as the Greenwood function only defines the pitch at the level of the BM. To examine the consequences of the length difference between OC and SG, the existing computational model has been enhanced to incorporate more realistic fibre trajectories that are based on experimental data.

A comparison of electrical field imaging (EFI) recordings (Vanpoucke et al., 2004), obtained intra-operatively in patients and simulated in the computational model, revealed that intra-cochlear potentials in the model were roughly a factor 10 lower than values measured in patients. However, many of the conductivities used in the model were estimated values, particularly the bone conductivity, which was based on measurements from the lateral wall in guinea pig cochleae (Suesserman, 1992) and might have a considerably different conductivity value than the human temporal bone it was intended to model. The conductivity value used for the modiolus is also debatable, as it is a complex structure that has been simplified to a homogeneous object in the model. Furthermore, the modiolus contains neural compartments which are anisotropic in nature, but which have been modelled with isotropic conductivity. Since the conductivities of bone and the modiolus are expected to have the largest impact on the amount of current leaking from the cochlea, the model has been upgraded by including fluid compartments inside the modiolus and by updating the tissue conductivities through use of clinical intra-cochlear recordings.

To assess the influence of anatomical variability, four basic human cochlear geometries were used, with implants modelled in both perimodiolar and lateral wall positions, deeply inserted to also study stimulation in the cochlear apex. The updated model was employed to investigate to what extent the Greenwood function can be used directly to estimate the pitch percept of CI stimulation, especially in the apex of the cochlea. Also, the amount of overlap between the excited neural subpopulations in the model was used to estimate the discriminability of different electrode contacts along the cochlear scalae. Furthermore, geometric data from the model was used to determine OC and SG length at a given cochlear position, in order to describe place pitch as a function of cochlear angle. The results may provide clinical insight into the optimal length and position of electrode arrays, and the functioning of electrode contacts near and beyond the end of the SG.

2. Materials and methods

The model used in this study is an extension of the previously published computational model of the electrically stimulated cochlea developed at the Leiden University Medical Centre (Briaire and Frijns, 2000a,b, 2005, 2006; Frijns et al., 2000, 2001). The volume conduction part of this model uses a realistic three-dimensional geometry representing a human cochlea implanted with a multi-channel electrode array. Electrical potentials at the Ranvier nodes of the neurons in this geometry are calculated using the Boundary Element Method, and then coupled to an active nerve fibre model to predict neural excitation.

2.1. Geometric changes

For the purpose of this study, the cochlear geometry has been updated in a number of areas, in an effort to more accurately model cochlear anatomy. Firstly the trajectories of the axons of the cochlear nerves have been altered to bundle up together in the modiolus, instead of running parallel to the modiolar axis from the cell bodies downwards (compare Fig. 1a and b). The diameter of the nerve bundle is roughly 1 mm, comparable to the diameter of a real life cochlear nerve. In addition to bundling up, the trajectory in the modiolus is curved, rather than extended in the direction of the rotational axis of the cochlea. The direction and slope of the curvature was estimated from a number of clinical CT-scans. Furthermore, a fluid sheath has been added around the cochlear nerve bundle, and a spiralling compartment of fluid was made between the nerve fibres inside the modiolus of the model geometry (outlined grey areas in the top images of Fig. 1). These fluid compartments are visible in histological cross-sections of the human