A 1-V Nanopower Highly Tunable Biquadratic $G_m - C$ Bandpass Filter for Fully Implantable Cochlear Implants

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**Abstract**—This paper presents a low power highly tunable second-order $G_m - C$ bandpass filter for auditory signal processing of fully implantable cochlear implants. The resonance frequency of the filter is tunable within the daily acoustic band of 200 - 6000 Hz with a tunable quality factor from 1 to 3. The operational transconductance amplifiers (OTAs) operate in the subthreshold region to achieve low power consumption. The input transistors are driven from the bulk to reduce the transconductance so that the low frequencies of the daily acoustic band are covered. The effect of reduced transconductance on the gain is compensated by increasing the output resistance with folded cascode structure so that unity gain is achieved. The filter was fabricated using TSMC 0.18-µm CMOS technology on an active area of 0.085 mm$^2$. The filter consumes 13.2 nW from a 1 V supply and achieves 51.98 dB dynamic range at the resonance frequency of 950 Hz.

**Index Terms**—Fully implantable cochlear implant, auditory signal processing, analog bandpass filter, low power consumption

**I. INTRODUCTION**

The hair cells in the inner ear are sensory cells that convert mechanical vibrations in the cochlear fluid into the electrical signals transmitted to the auditory nerve. Sensorineural hearing loss occurs when the hair cells are damaged or lost. Cochlear implant are promising solutions for the treatment of sensorineural hearing loss. The operational principle of the cochlear implants is based on the stimulation of the auditory neurons in the cochlea artificially. As the nerve endings are stimulated directly, the damaged hair cells together with the eardrum and the ossicles are bypassed. In general, conventional cochlear implants consist of both external and internal units. In the external unit, the audio signals are picked up by the microphone and processed by the speech processor. The processed signal is then transmitted to the internal unit through an RF link where the electrical signals to stimulate the auditory nerve are generated. On the other hand, fully implantable cochlear implant (FICI) systems have the potential to eliminate the external components of conventional cochlear implants and realize the same operation in the middle ear.

The sound waves are sensed by an implanted acoustic transducer and the converted signal is transmitted to the speech processor in fully implantable cochlear implants. To have a high quality of hearing, the speech processor should be able to interpret different frequency content of the audio signal of interest effectively, which is investigated in this study. These frequency bands are typically determined by the electrode array of cochlear implant and typically around 8 to 16 center frequencies (100-8000 Hz) with a quality factor of 1 to 6. The speech processor then extracts the envelope of each filtered signal and the amplitude of the output is modulated to adapt each signal to the perceptual threshold of the patient. In the end, the processor generates the stimuli for neural stimulation proportional to the energy of the input signal and sends it to the cochlear electrodes [1]. The operation of the speech processor and the focus of this research is summarized in Fig. 1.

This work introduces a low power highly tunable biquadratic $G_m - C$ bandpass filter to mimic the decomposition of a sound in the cochlea. To cover the daily acoustic band, the input transistors of OTA cells are driven from the bulk as the gate-transconductance is typically three to five times smaller than gate-transconductance. To reduce the power consumption and extend the battery life, the OTA cells are set to operate in the subthreshold region. To maximize the performance of the implant for patient fitting, the filter topology is set such that the resonance frequency, bandwidth, and gain are tunable through the bias currents and capacitors. The combination of this work with a single channel non-resonance mode piezoelectric acoustic transducer has the potential to result in a novel fully...
implantable cochlear implant system. Unlike common middle ear implants that actuate the ossicles through their transducer mounted onto incus, our approach is based on not actuating the ossicles but sensing the vibration of the ossicles, processing the signal through the proposed filter, and stimulating the cochlea. The designed OTA cell and filter are briefly described in Section II. Section III presents the experimental results of the chip. Finally, the work is concluded in Section IV.

II. Circuit Design and Analysis

A. Folded Cascode Transconductor Cell

The operational transconductance amplifier is the main building block of the proposed $G_{m} - C$ filter. The frequency range 250 Hz - 6500 Hz is stated as the ideal frequency band for the perception of the speech signal by a cochlear implant [2]. Therefore, the transconductance of the input transistors should be low enough to cover this frequency band. On the other hand, the power consumption of the OTA cell should be as low as possible to extend the battery life of the cochlear implant. Based on these considerations, a bulk-driven folded cascode OTA that operates in weak inversion is proposed in this research as shown in Fig. 2. In this structure, the transistors M7 and M8 act as PMOS differential input pairs. The input bias current is mirrored through the PMOS cascade stage to the output biasing branch and the source of the input pair. The M9-M10 provides the DC bias voltage to the transistors M11-M14 form the cascode load configuration to increase the output resistance for the gain. All transistors are designed to operate in weak inversion region for low power consumption. The drain to source voltages are set to be larger than four times the thermal voltage. The output resistance and gain of the proposed OTA are given by the following expressions:

$$R_{out} = r_{oM16}(g_{mM16}[r_{oM16}][r_{oM14}])[r_{oM14}(g_{mM14}r_{oM14})]$$  (1)

$$A_{v} = g_{mM16}[r_{oM16}(g_{mM16}[r_{oM16}][r_{oM14}])][r_{oM14}(g_{mM14}r_{oM14})]$$  (2)

where $g_{mbo}$ refers to bulk-to-source transconductance and $g_{m}$ refers to gate-to-source transconductance.

To cover low frequencies of the daily acoustic band, $g_{m}$ values should be small enough. However, the reduction in the $g_{m}$ value to achieve low frequencies causes the reduction in the gain as (2) suggests. Therefore, the folded cascode structure is utilized to increase the output resistance. In other words, the effect of reduced $g_{m}$ on the gain is compensated by increasing the output resistance. To lower the input transconductance, the input transistors are driven from the bulk since the bulk-transconductance is three to five times smaller than gate-transconductance. In this technique, the gates of M7 and M8 are grounded to ensure the inversion layer is formed between the drain and source terminals. The current flow through the transistors is controlled by the threshold voltage change which is a function of bulk-source voltage. The reduction in the transconductance is expressed from the subthreshold current (3) and threshold expressions (4) as follows:

$$I_{D} = I_{D0} \frac{W}{L} e^{\frac{V_{GS}-V_{TH}}{nV_{t}}} (1 - e^{-\frac{V_{DS}}{nV_{t}}})$$  (3)

$$V_{TH} = V_{T0} + \gamma (\sqrt{|2 \phi_{f}|} + |V_{SB}| - \sqrt{|2 \phi_{f}|})$$  (4)

where $\gamma$ is the Body effect coefficient and $\phi_{f}$ is the Fermi level potential.

The gate (5) and bulk (6) transconductance are defined as follows:

$$g_{m} = \frac{\partial i_{D}}{\partial v_{GS}} = \frac{I_{D}}{nV_{i}} = - \frac{\partial i_{D}}{\partial v_{TH}}$$  (5)

$$g_{mb} = \frac{\partial i_{D}}{\partial v_{BS}} = \frac{\partial i_{D}}{\partial v_{TH}} \cdot \frac{\partial v_{TH}}{\partial v_{BS}}$$  (6)

The terms in the extension of (6) are found as follows:

$$\frac{\partial i_{D}}{\partial v_{TH}} = -g_{m} \frac{\partial v_{TH}}{v_{BS}} = \frac{\gamma}{2 \sqrt{|2 \phi_{f}|} + |V_{SB}|}$$  (7)

The derivative of threshold voltage with respect to bulk-to-source voltage in (7) can be assumed as a constant value that varies from 0.2 to 0.4. Then, from (6), the bulk transconductance is calculated as three to five times lower than the gate transconductance [3].

B. Biquadratic $G_{m} - C$ Band Pass Filter

The optimum filter order and quality factor of a cochlear filter bank with 6 to 12 channels are specified as $N=2$ and $Q=4$ for the ideal speech perception [4]. In our application, it is preferred to shift the frequency range indicated in [2]...
to 200 - 6000 Hz to cover male/female voice frequencies with 8 cochlear channels better. In addition, apart from [4], it is preferred to have a quality factor 1-3 to apply Advanced Combination Encoder (ACE) strategy [5] effectively. Last but not least, the resonance frequencies and the quality factors should not be fixed but adjustable by an audiologist for the patient fitting so that the speech perception would be maximized for each individual. Therefore, a highly tunable biquadratic \( G_m - C \) bandpass filter is proposed in this research as shown in Fig. 3. The structure consists of four OTA cells and two capacitors which act as a resonant RLC filter. The second and the fourth OTA cell with \( C_2 \) form an active inductor and set the resonance frequency of the filter with \( C_1 \). The third OTA cell acts as an active resistor and reduces the resonance of the filter. In addition, it determines the output DC level as \( V_{ref} = \frac{V_{DD}}{2} \) through the negative feedback. The first OTA cell is an attenuator and compensates for the non-ideal effects on the gain. The transfer function of the filter is defined as follows:

\[
H(s) = \frac{G_{m1} \cdot s}{s^2 + \frac{G_{m2}}{C_1} s + \frac{G_{m3} \cdot G_{m4}}{C_1 C_2}} \tag{8}
\]

The center frequency, quality factor and gain of the proposed filter can be extracted from (8) as follows:

\[
\omega_0 = \sqrt{\frac{G_{m2} \cdot G_{m4}}{C_1 C_2}}, \quad Q = \frac{1}{G_{m3}} \sqrt{\frac{G_{m2} \cdot G_{m4} C_1}{C_2}}, \quad H_0 = \frac{G_{m1}}{G_{m3}} \tag{9}
\]

These expressions (9) summarize the tunability of the filter. To set the center frequency, the transconductances of the OTA2 and OTA4 or the capacitances of \( C_1 \) and \( C_2 \) can be altered. To set the quality factor of the filter, either the transconductance of the OTA2, OTA3, and OTA4 or the capacitance of \( C_1 \) and \( C_2 \) can be adjusted. To set the gain, the ratio between the transconductances of the OTA1 and OTA3 can be changed. The transconductances of the cells can be controlled through the bias currents. In this research, the adjustments on the center frequency and quality factor are realized through the bias currents and capacitors while the gain tuning is realized through the bias currents of the cells. To further reduce the power consumption at low frequencies, an additional capacitor is added to the output node during the chip tests which modifies the value of \( C_1 \). On the other hand, there is no independent parameter which would be the drawback of the structure. In other words, although it is possible to keep the resonance frequency constant while altering the quality factor through the transconductance of OTA3, this adjustment causes a variation in the gain as well. This effect would be compensated at the preamplifier stage.

### III. Experimental Results

The proposed circuit was implemented in TSMC 0.18-µm CMOS technology. Fig. 4 shows the experimental setup of the system and Fig. 5 presents the micrograph of the filter with an active area of 0.085 mm\(^2\). In addition to the filter, there is an observation buffer to eliminate capacitive loading from the oscilloscope probes.

The in-band linearity performance of the filter is measured with 1000 Hz sine wave. 1 dB compression point is observed at -21.7 dBV as indicated in Fig. 6. This result is in good agreement with the simulation result of -21.5 dBV. As this work is intended to be used with a non-resonance mode piezoelectric acoustic transducer, the distortion level or linear range around these values are acceptable since the output level of the non-resonance transducer will be well below these values. The noise is integrated within the passband of the filter and the rms value of the input-referred noise is calculated as 207 µV which was 179 µV in the simulations. As a result, the dynamic range (DR) of the filter is calculated to be 51.98 dB.
The frequency response of the filter is obtained through a sinusoidal input signal of 250 mV \text{pp} under 1 V supply. In our research, it is intended to cover the daily acoustic band of 200 Hz-6000 Hz with eight center frequencies which correspond to a cochlear electrode with eight nodes as shown in Fig. 7(a). The center frequencies are selected such that it is distributed linearly below 1000 Hz and logarithmic above 1000 Hz based on ACE strategy. As a result of the logarithmic increase, the quality factors of the resonances above 1000 Hz are decreased slightly compared to resonances below 1000 Hz to provide better coverage of the daily acoustic band. Therefore, the quality factors vary from 1.3 to 1.8 in this result as Fig. 7(b) presents which is almost ideal for our application of 8 channel fully implantable cochlear implant. However, for the applications of cochlear implants with electrode nodes higher than eight, it is possible to adjust the quality factor of the filter between 1 to 3. The gain during the chip test is in between -1.4 dB and -1.6 dB. However, it is possible to change the gain depending on the application as shown in Fig. 7(c). These results summarize the tunability performance of the proposed filter. By adjusting bias currents to transconductance amplifiers and switching the capacitors, the center frequency of the filter can be adjusted from 200 Hz to 6000 Hz and the quality factor of the filter can be varied from 1 to 3. Even if several nodes of the cochlear electrode were not successfully implanted, it is possible to rearrange the resonance frequencies and bandwidths to recover the lost band after the implantation thanks to the proposed filter. For this application, the power consumption of the filter is the most important parameter to extend the battery life of the cochlear implant and the proposed filter consumes only 13.2 nW at 950 Hz under 1 V supply voltage.

The resonance frequency of the filter would be different than the value set during the design after the fabrication due to process variations and the device mismatch. Fig. 8 shows the Monte Carlo simulation result of the resonance frequency set to 950 Hz over 200 runs. An average resonance frequency of 951 Hz is obtained with a standard deviation of 47 Hz. The possible shifts from the intended resonance frequency after the fabrication would be compensated by adjusting the bias currents of the second and fourth OTA cells so that the desired frequency response would be achieved.

Finally, Table I compares the performance of the proposed filter to the CMOS bandpass filters that have been reported in the literature. This comparison table only considers the works with experimental results excluding the works with simulation results. The power consumption, dynamic range, and coverage...
of daily acoustic band are the important parameters for a fully implantable cochlear implant filter which define the filter Figure-of-Merit as [7]:

$$FOM = \frac{Power \cdot V_{DD}}{N \cdot f_0 \cdot DR}$$

(10)

where $N$ is the filter order, $DR$ is the dynamic range and $f_0$ is the center frequency. The lower FOM indicates a better performance.

Table I confirms that the proposed filter achieves a wide tuning range that covers the daily acoustic band while consuming power in the order of nW; resulting in a lower (hence better) FOM compared to the other highlighted designs.

IV. CONCLUSION

In this paper, a low power 2nd order $G_m - C$ bandpass filter with tunable resonance frequency, bandwidth, and gain has been presented. The wide tuning range of the filter within the daily acoustic band (200–6000 Hz) with a tunable quality factor (1-3) makes the proposed filter ideal for the auditory signal processing of fully implantable cochlear implants. The low power consumption (13.2 nW @ 950 Hz) of the structure has a promising future to further decrease the power consumption of analog front-end so that the battery life of fully implantable cochlear implants would be extended. The experimental results show that the dynamic range (51.98 dB) of the proposed filter is compatible with the non-resonance mode piezoelectric acoustic transducers. The filter will be tested with a properly designed single channel non-resonance mode piezoelectric acoustic transducer as a part of a novel fully implantable cochlear implant system.

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