Hearing instruments go digital.

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Abstract.

This paper describes the backgrounds, required performances and choices made at system and circuit level to make a digital hearing aid, competitive in power and area with analog solutions. Power consumption of the digital circuitry working at 1.1 V equals that of the on-board analog interface circuitry. Though chip area is 35 square mm, and digital area takes about 80 % , the rapid development of CMOS processes makes this the turning point in going from full analog to digital solutions.

Introduction.

The addiction of nowadays youngsters to music, played at very high loudness levels, makes the future for manufacturers of hearing instruments look very bright. At present about 12% of the human population suffers from hearing problems. About 18% of those are suffering relevant perceptual loss, from which a few percent actually use hearing instruments. Present day functionality is to adapt the incoming audio signal both in frequency response and dynamic range to a damaged human ear. To adapt the device to the individual, a solution with large flexibility is required. Up to now the required adaptations have been limited to splitting the audio frequency range in a number of frequency bands, adapting the frequency response, and applying compression per band. More complex functionality will be needed in future, speed of innovation depending upon better understanding of the patients needs, and the availability of powerful flexible programmable devices. Digital signal processing offers superior flexibility, but until now at the cost of a higher power dissipation and more chip area. As the chip size is limited by the size of the human ear, only advanced technology will make this feasible. Future reduction of area and power in the analog domain is limited by noise and accuracy requirements. Several digital hearing aid IC's have been reported so far, [1],[2], and major players in this market have already introduced digital hearing instruments with limited programmability. A device, introduced by Philips recently, offers full programmability by the audiologist. Such a device is a real single-chip system, controlled via an on board infrared receiver. By using a dedicaded low-threshold process, low voltage digital libraries and power saving DSP structures, the digital core operates at 1.1V. Stabilised power supplies are implemented by both classic regulator as well as charge pump configurations. Total power consumption is 2 mW with a 35 mm2 chip area. These results were obtained using a 0.8 micron process. It is obvious that future developments in more advanced processes will be digital.

Some history.

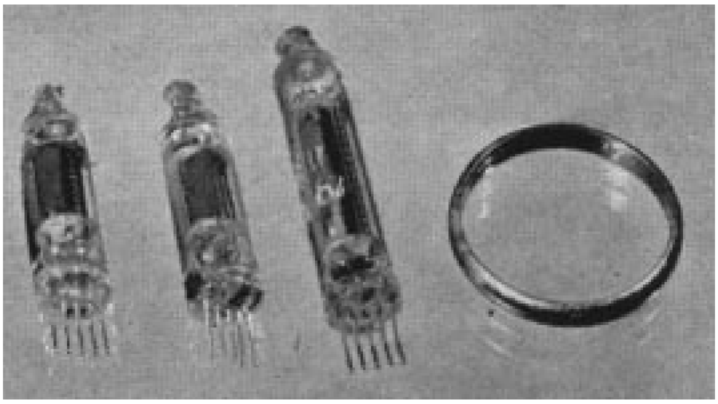


Fig. 1. Three amplifier valves, with a wedding ring for comparison of size.

It is not known at what time in history human beings started to use mechanical devices to improve their audible perception. The first method was without doubt to use the hand. All other methods are more expensive, less reliable and less comfortable. Electronic signal processing to compensate for loss of perception, became feasible after the introduction of lightweight, battery operated amplifiers. Portable equipment was already produced around 1950 using miniature tubes. [3], Fig.1.The circuits were relatively simple as shown in Fig. 3 and though construction was compact it still was a box as shown in Fig. 2, carried around in the pocket. The introduction of transistors meant a breakthrough both in size and power consumption.

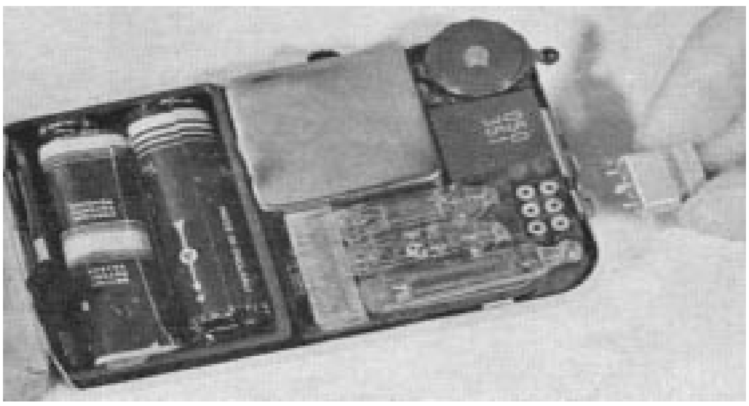


Fig. 2. Interior of a hearing aid with miniature tubes. Only the screened microphone, volume control, valves, earphone plug and batteries can be seen.

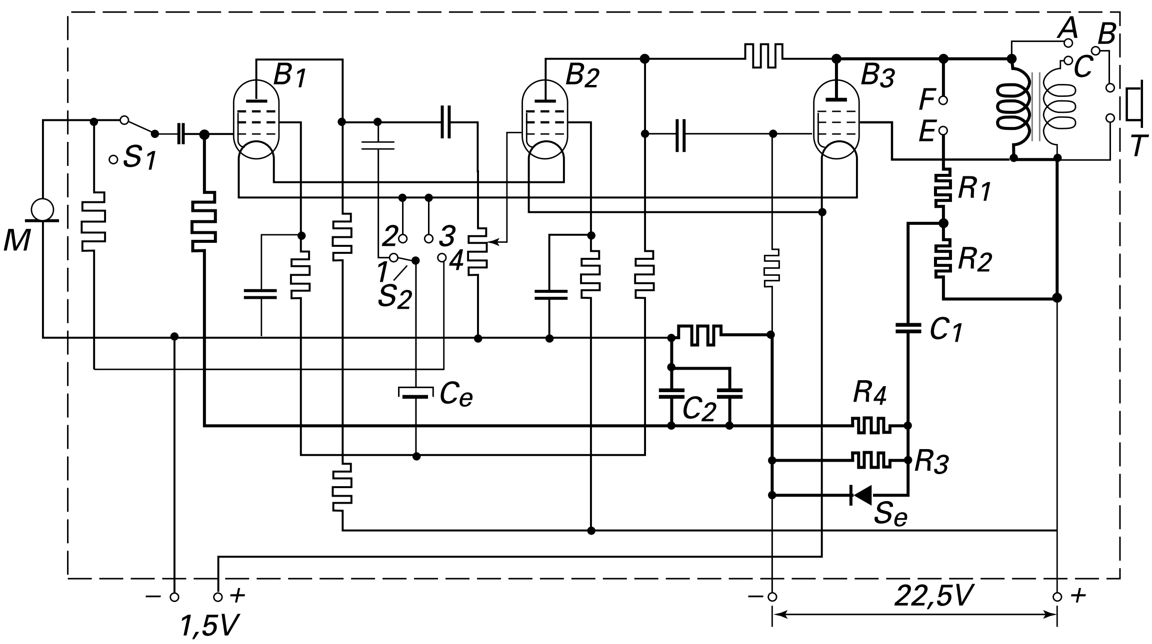


Fig. 3. Circuit diagram of the amplifier. The rectifier diode provides compression with asymmetric attack and decay times.

A remarkable first ’IC ’ prototype was made in germanium with diode and air isolation obtained by mechanical sawing around 1961. The circuit diagram of Fig. 4 can be recognised in the ’layout’ and construction of Fig. 5. Total assembly is shown in Fig. 6. The introduction of this very simple three transistor amplifier in silicon planair IC technology, called the OM 200, brought again strong reduction in size. Actually this device is stil in production today.

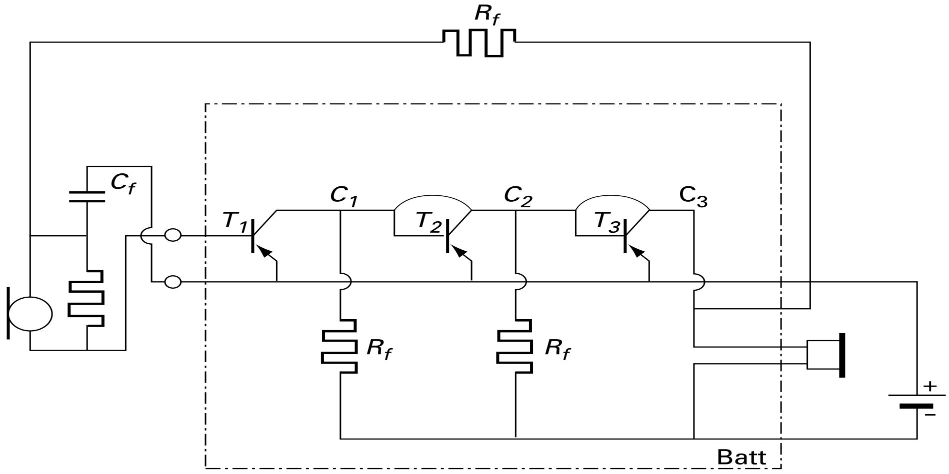
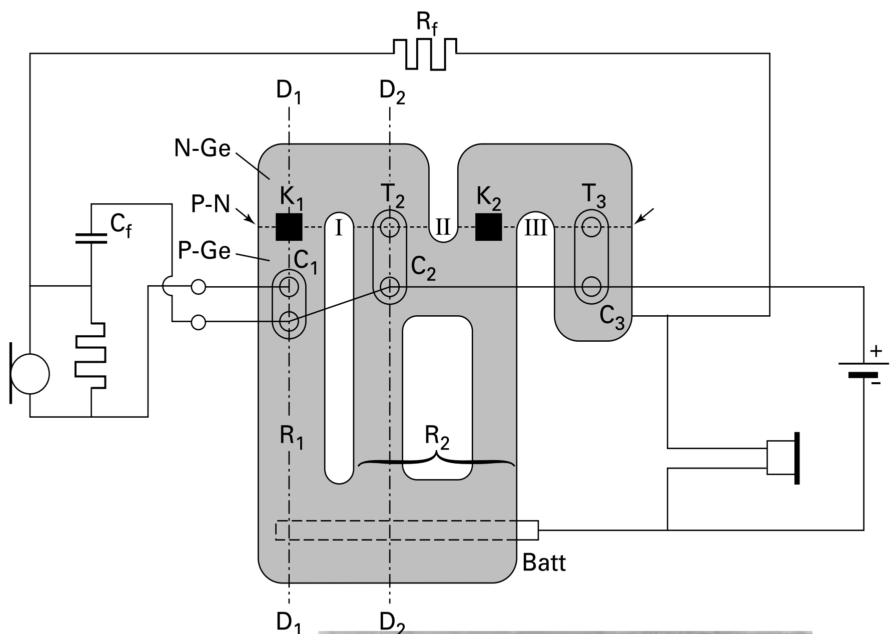
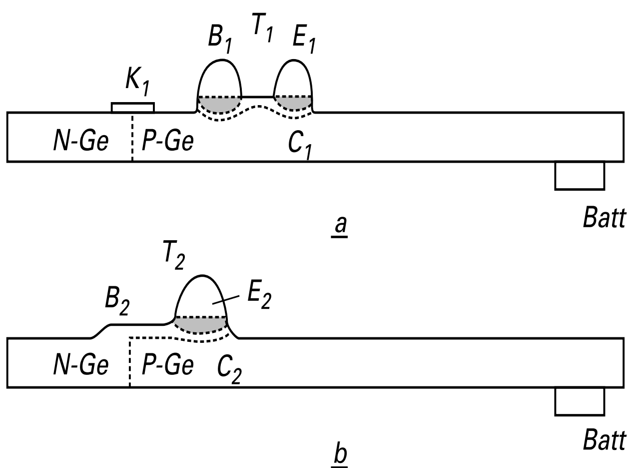


Fig. 4. Circuit diagram of the prototype germanium "integrated" amplifier.

Fig. 5. Basic transistor structure, "lay-out" implementation and photograph of the sawn device.





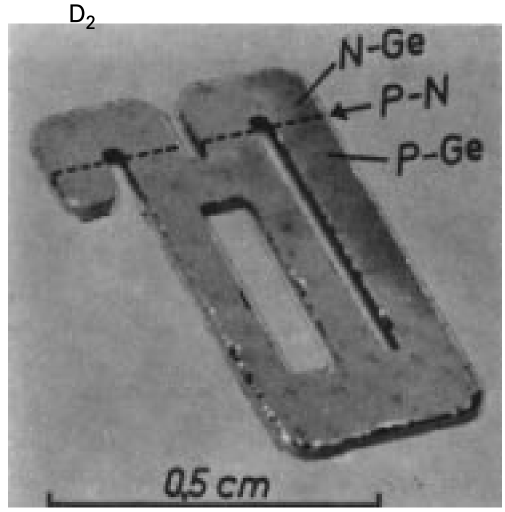
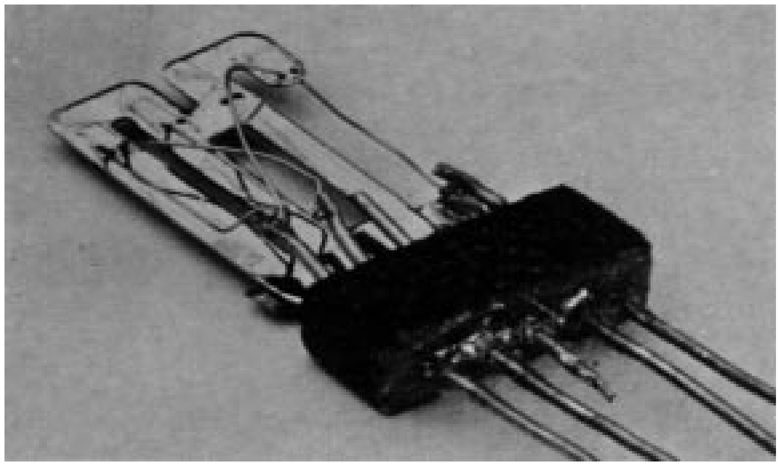


Fig. 6. Experimental version of the three stage amplifier. Bond wires, going to a common emitter lead in the circuit mounting have been attached to each of the emitter beads.



Perception background.

Everybody who has ever tried to retrieve information from the sound recorded by a microphone in a noisy environment knows that amplification only is not a satisfying method. So the first question will be: When has the audio perception of a 'normal' human degraded to such an extent that it will improve when using a hearing instrument. Secondly: What signal processing and performance levels are needed from the hearing instrument ? The first question can partly be answered by test material gathered in the past.

The classical graphs on human perception are given by fig. 7, which shows the equal loudness contours as well as the minimum audible field curve as a dotted line. In audiometry mostly the MAF curve is taken as a reference, and the equal loudness curves are drawn as differences to the MAF curve.

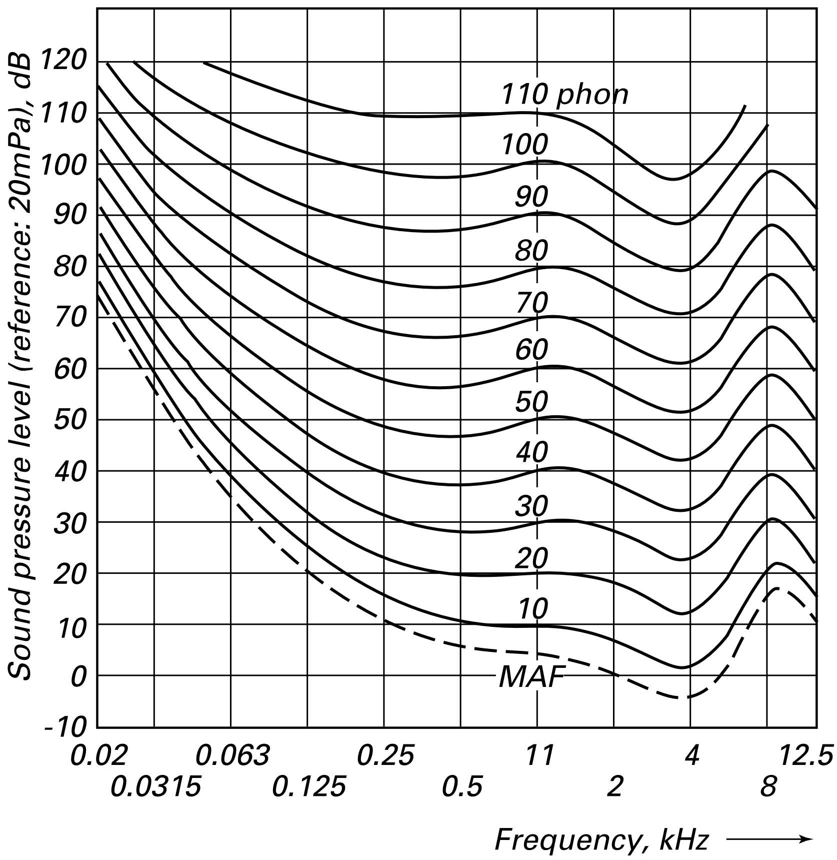


Fig. 7. Normal equal-loudness contours for pure tones (binaural free-field listening, frontal incidence).

A damaged ear however may show the curves of Fig. 8, where both frequency compensation and frequency dependent compression are needed. Fig. 8 also shows that correction is limited by maximum sound pressure

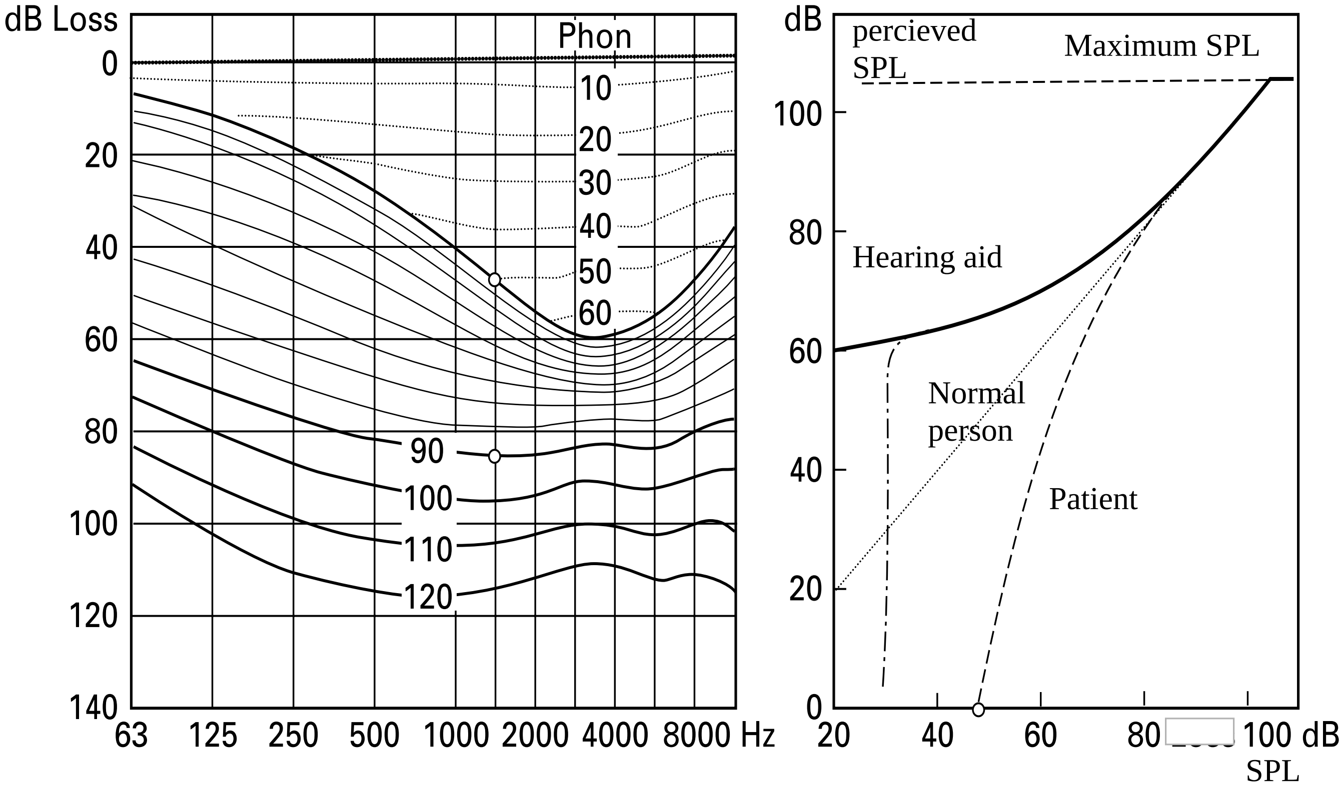


Fig. 8. Normal equal-loudness contours for pure tones (binaural free-field listening, frontal incidence).

level on one side, and a maximum gain level on the other side. This maximum gain level is partly dependent on the sophistication of the hearing device. Also too much amplification increases the background noise level, which increases the hearing threshold. Another limit can be acoustical feedback, specially for small devices to be used in the ear. Actually the right correction characteristic is not only depending on the ear, but also on the type of sounds received. So there is a need for very complex signal processing, which is not actually well-defined, as it depends on human interaction.

Market situation.

World population is about 5 billion people, of which about 2 % are in need of a hearing device. Only a small percentage actually uses such devices which makes the world market about 5 million pieces a year. A number of companies are serving this market with each a few hundred thousand pieces. This is not a very bright situation to allow for large investments such as the development of complex dedicated IC's. Is it likely that the future market will grow due to loud music for instance? A survey of tests done around 1980 among visitors of concerts and musicians, as well as visitors of disco's, of which material is shown in Fig.9 and 10, showed little detoriation of perception due to loud music, with the remark that most data were obtained from listeners subjected to SPL's of 110 dB or less.[4],[5],[6]. Recent investigation of disco's in Poland , published at a workshop on Audio technology for the hearing impaired, during an A.E.S. conference in Munich 1997, showed SPL 's of 130 dB or more. Those levels are comparable with industrial noise which has proven to cause severe degradation of perception. As both industrial and leisure sound levels are obviously without proper government control in many parts of the world, we can predict that total market percentage indeed will increase.

Sound levels in discotheques 1970-79					
AUTHORS	YEAR	PLACE	COMMENTS	LIN	dB(A)
Abrol, Nath Sahai	1970	New Delhi	Original paper inaccessible; data taken from Whittle & Robinson	89	84
				88	83
				100	95
				94	88
				83	79
				93	89
				89	88
				92	90
Fearn	1972	UK	Room centre	94	92
			Near Loudspeaker	108	100
Rupp, Banachowski, Kiselwich	1974	USA	Means of many rdgs.		100
Sherreffs	1974	USA	Means of several rdgs.		107
Cabot, Genter Lucke	1974	USA	Disco 02: 4 times	98	95
				96	91
				93	90
				93	87
				94	82
			Disco 15: once	94	87
			Disco 17: 4 times	95	92
				94	86
				94	86
Bickerdike & Gregory	1980	USA	Disco 11: unlicensed	108	101
				110	105
			Disco 18: licensed	110	103
				108	102
			Disco 29: licensed	106	99
				109	101
			Disco 30:licensed	117	113
				117	113
			Disco 31: licensed	94	88
				95	87

Fig. 9. Taken from [4], survey of sound levels in discotheques.

Hearing thresholds of various groups of otologically normal young people								
AGES	N	dB HL 500Hz	1 kHz	2 kHz	3 kHz	4 kHz	6 kHz	8 kHz
(a) Those who do not attend discotheques/pop concerts								
9-12	83	4.2	3.0	0.8	1.3	2.1	3.8	2.5
13-16	135	6.8	1.3	0.9	0.1	2.4	6.8	5.3
(b) Those who attend discotheques/pop concerts (same Sources)								
9-12	61	7.9	4.8	1.6	2.8	4.8	8.1	7.2
13-16	88	8.2	3.8	1.7	2.0	4.2	7.9	7.0
(c) Differences between above two sets of data								
9-12		3.7	1.8	0.8	1.5	2.7	4.3	4.7
13-16		1.4	2.5	0.8	1.9	1.8	1.1	1.7

Fig. 10. Taken from [4], an adaption from [5] and [6], showing a slight increase of hearing loss for youngsters,exposed to sound levels comparable to Fig.9.

Technology push.

A number of companies have recently introduced devices based upon digital signal processing rather than the conventional analog approach. The main reason for going digital is flexibility and programmability. Not only frequency compensation and dynamic range adaption have to be fine-tuned to the patient needs, but should also be dependent on the type of received audio signal. Todays CMOS technology brings DSP solutions within the power and size budget, even for in-ear solutions. Single battery cell operation and minimum power consumption is of prime importance here. To explore this flexibility, the device must be field programmable and have some sort of remote control. This can be solved by using an IR link both for control and to download programs. A block diagram of such a device is given in Fig. 11. We will now focus on some electronic constraints regarding such developments and circuitry to fullfil these constraints.

Dynamic range.

Input dynamic range of a hearing aid is typically 80 dB, while the microphone output signal is only 14 mV. Input noise level decreases with the square root of impedance while the impedance of active devices decreases linear with increasing current. So the input configuration of the A/D converter should be chosen to have maximum gain at the input which reduces noise contribution of the next stages. The most popular choise for A/D conversion, the Delta-Sigma converter acts as a type of feedback amplifier, with a large loopgain at low frequencies, which makes the input stage the dominant noise-source. Its high oversampling ratio allows replacement of analog anti-aliasing filters by digital filters. By using higher order loopfilters, a trade-off can be made between oversampling ratio and speed as shown in Table 1.

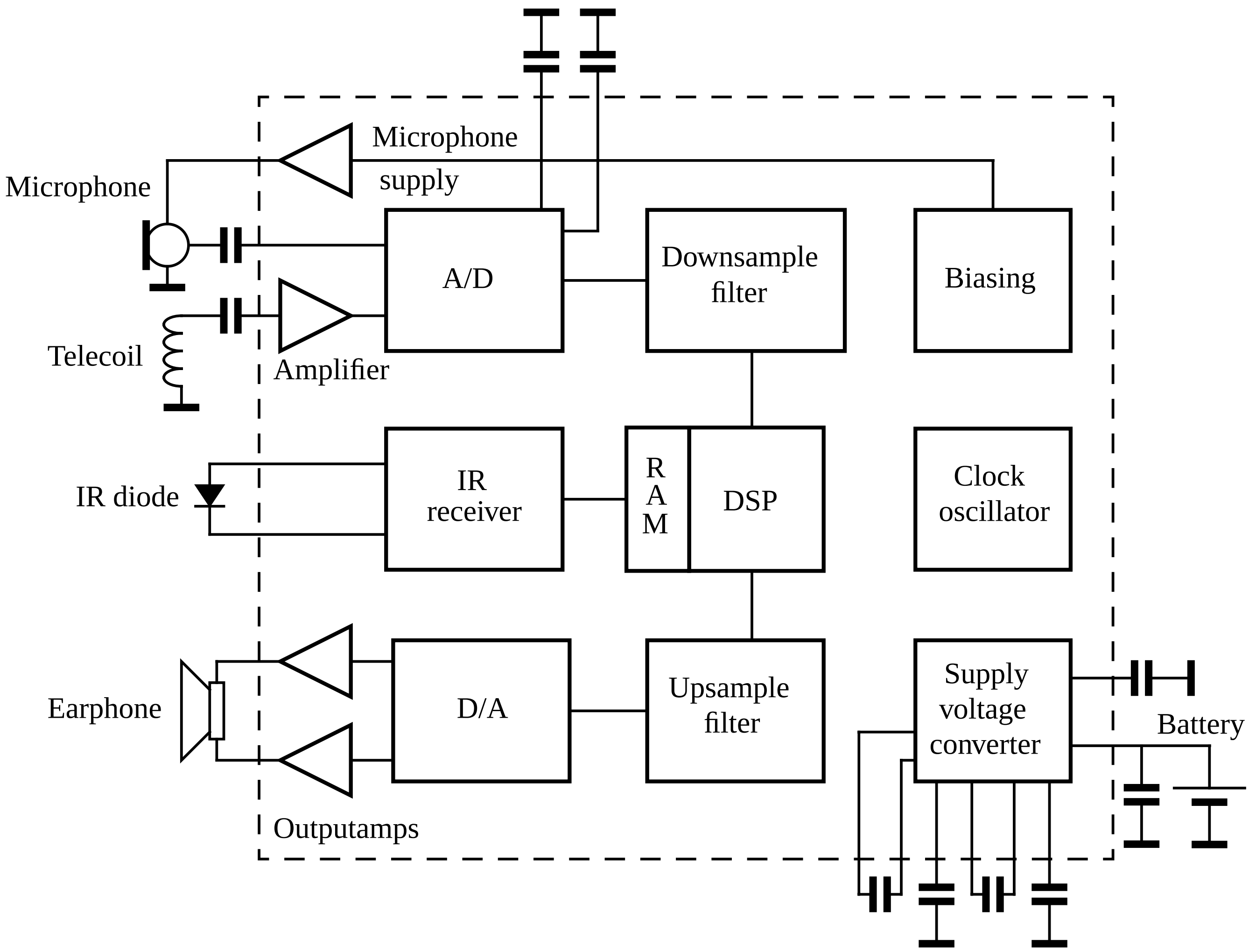


Fig. 11. Block diagram of the Hearing Aid IC.

order	m	overload level
2	128	-2 dB
3	75	-3 dB
4	59	-3 dB
5	58	-4 dB

Table 1: Loopfilter order and oversampling ratio for S/N_Q=90dB (real poles)

Related to power consumption it is better to reduce quantisation noise than thermal noise so the latter is chosen to be dominant. A continuous-time loop-filter of the modulator further decreases sensitivity for spurious input signals.[7]. When using Gm/C filters, the input amplifier can be incorporated in the input stage as shown in Fig. 12 .

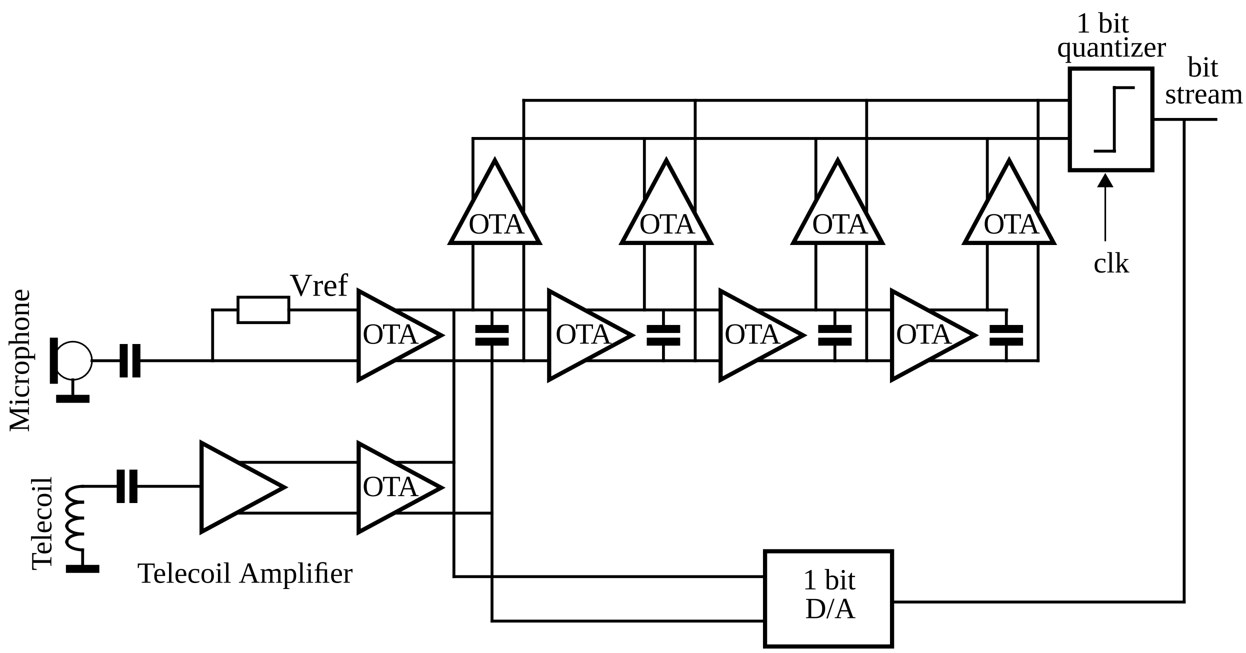


Fig. 12. Analog frontend with time continuous A/D converter

A part of the output load current sources of this amplifier serves also as the feedback D/A converter, shown in Fig.14, so a minimum amount of current is waisted.

Of course linearity of the input amplifier is dependent on the bias current, which is shown in the graph of Fig.13.

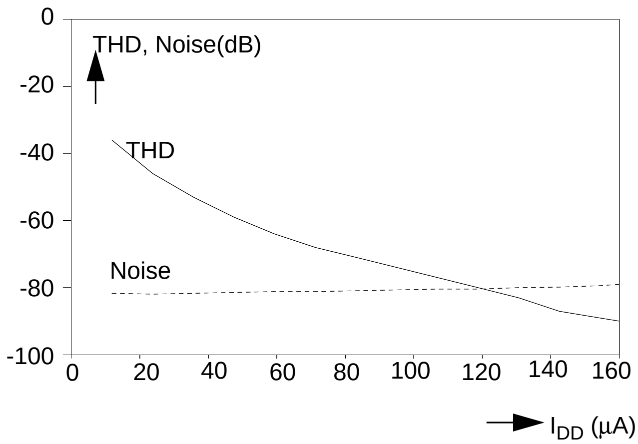


Fig. 13. Linearity and noise vs. current

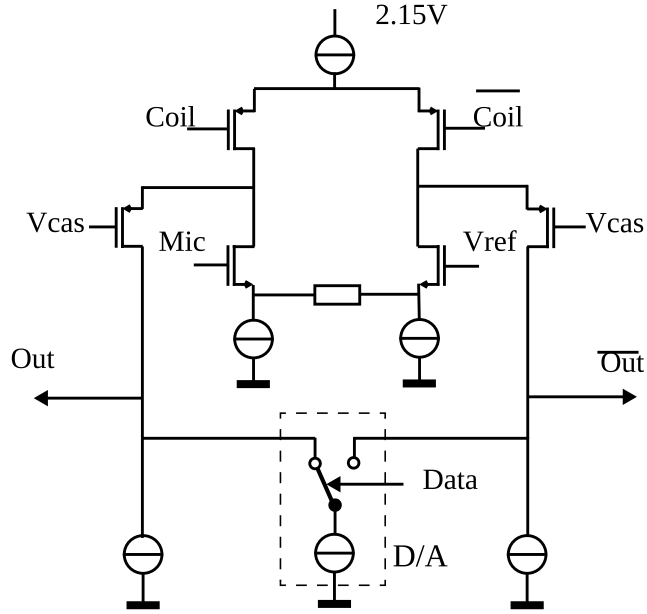


Fig. 14. A/D input stage

Power supply.

A small single battery cell has a voltage varying from 1.6 V in full condition to about 1.1 V in empty condition with a fairly constant 1.3 V during midlife. AC impedance varies from 10 Ohm full to 60 Ohm in empty condition. This behaviour determines the power supply strategy. DCDC conversion can only be implemented with (external) capacitors due to size. These converters are only efficient if the output voltage is an integer times the battery voltage or a binary division of the battery voltage, in which case efficiency can be about 90 %. For the digital part supply voltage had to be at least 1.1 V to obtain the required speed. A regulator for the digital supply implemented as a Delta modulator has been used here, shown in Fig.16, with a resistive switch acting as feedback resistor, and a stepsize control switching to bootstrapped charge-pump feedback when the battery voltage drops below 1.1 V.

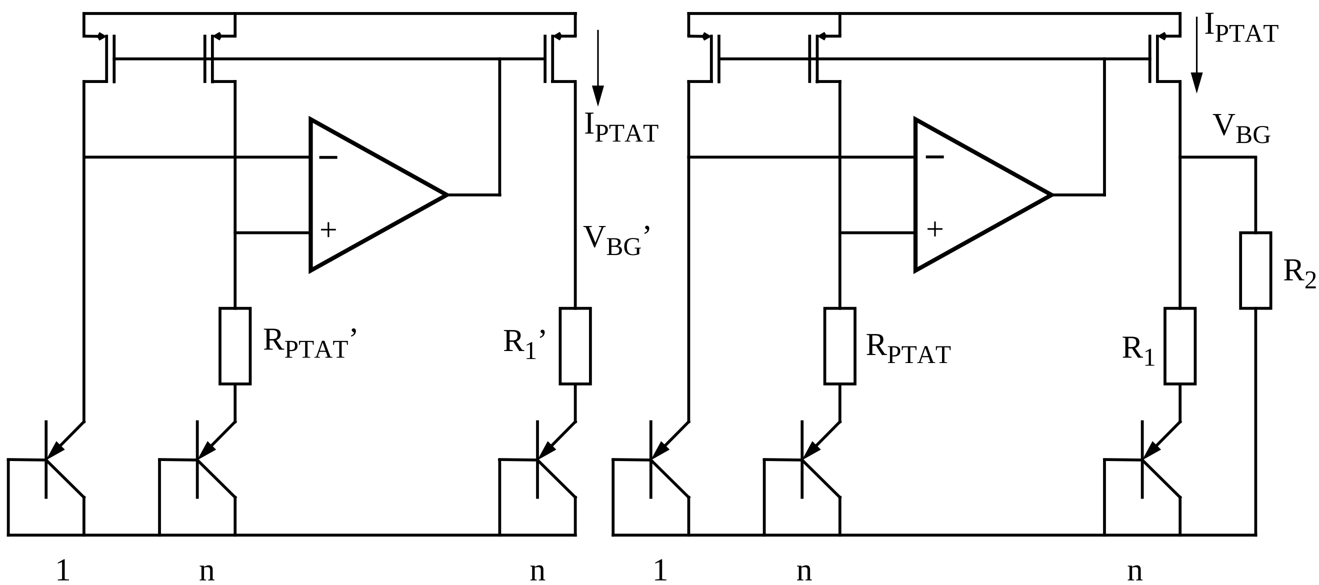


Fig. 15. Conventional and low supply voltage bandgap circuits

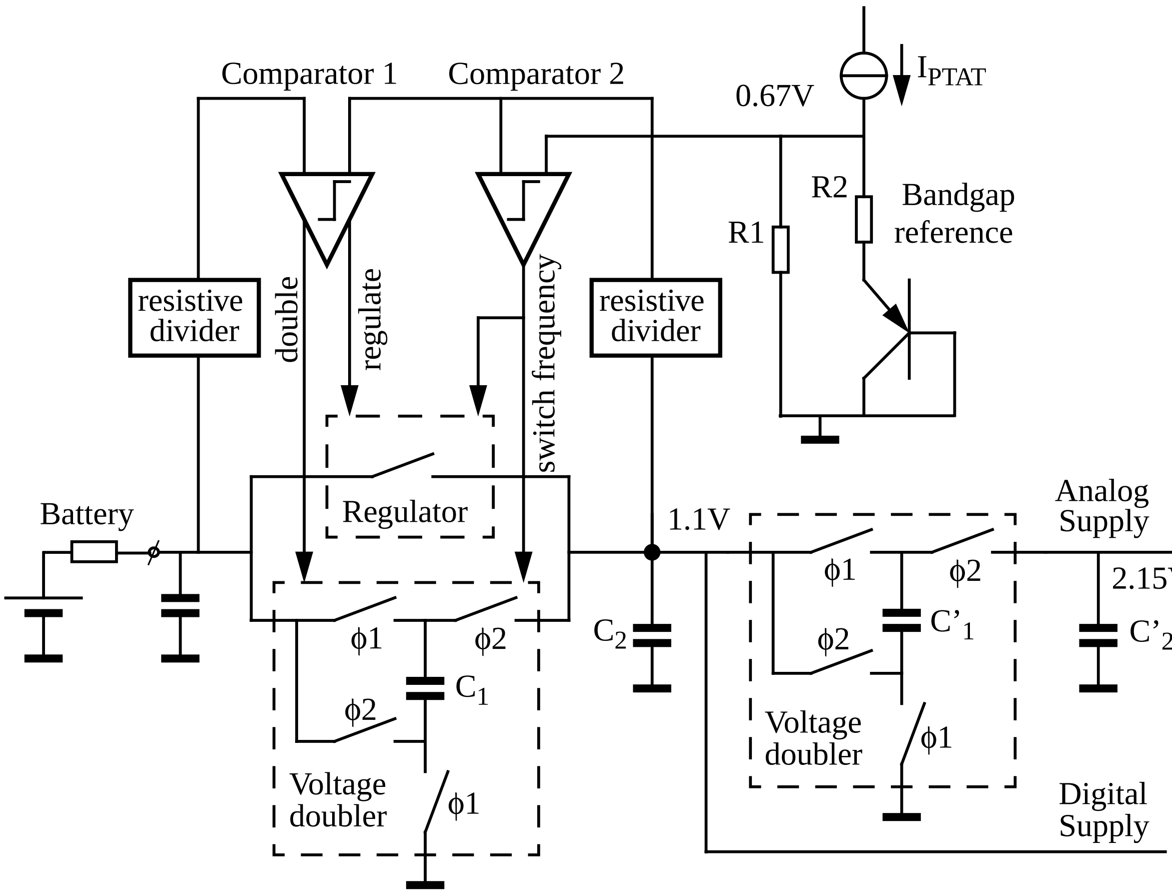


Fig. 16. Supply Voltage Converter

Simple voltage doubling from the 1.1 V is used then to generate an analog supply voltage of 2.1 V. A stable reference voltage is needed which can operate on 0.9 V supply voltage. The basic circuit is shown in Fig. 15 . A standard bandgap reference voltage is obtained when a current which is proportional with absolute temperature flows through a bipolar diode and resistor string. This voltage source can be seen as a temperature independent voltage source , with an output impedance given by the resistor and the diode impedance. When loaded with a resistor, the theoretical output voltage of 1.2 V will drop to an acceptable value of 0.67 V, but stays constant. The current through the load resistor is independent of temperature, while the the current through the diode changes more than proportional, depending on the current ratio.

Power supply rejection.

At maximum gain of 60 dB a square-wave output signal can be delivered to a 200 Ohm load. Worst case battery impedance is about 60 Ohms. Since large decoupling capacitors are not available, power supply rejection has to be at least better than 45 dB to avoid oscillation. It is much easier to obtain rejection by having a regulated power supply for the A/D converter. This also allows for some freedom of choice for the supply voltage. Noise voltages of active devices decrease with increasing current and analog voltage swings are limited by supply voltage minus fixed thresholds. Minimum power consumption for a given S/N ratio will be obtained when using maximum voltage swing,with high supply voltages, limited by maximum impedance level allowed for the required bandwidth. The optimum analog supply voltage for every stage is then determined by the voltage swing. About 40 dB rejection can be obtained from the regulator, and another 60 dB from the microphone supply and A/D input stage.

D/A converter.

The D/A converter system consists of the converter, output filters and output amplifiers. Again for high resolution power is saved when replacing analog output filtering by a moderate eigth times digital upsampling, filtering and first-order noise-shaping. Together with a 13 bit resistor string D/A convertor which generates two balanced outputs using only one current. Non-inverting output amplifiers show the necessary high impedance load for the D/A converter. The required output noise level of -93 dB is a sum of both quantisation and thermal noise sources as shown in Fig. 17

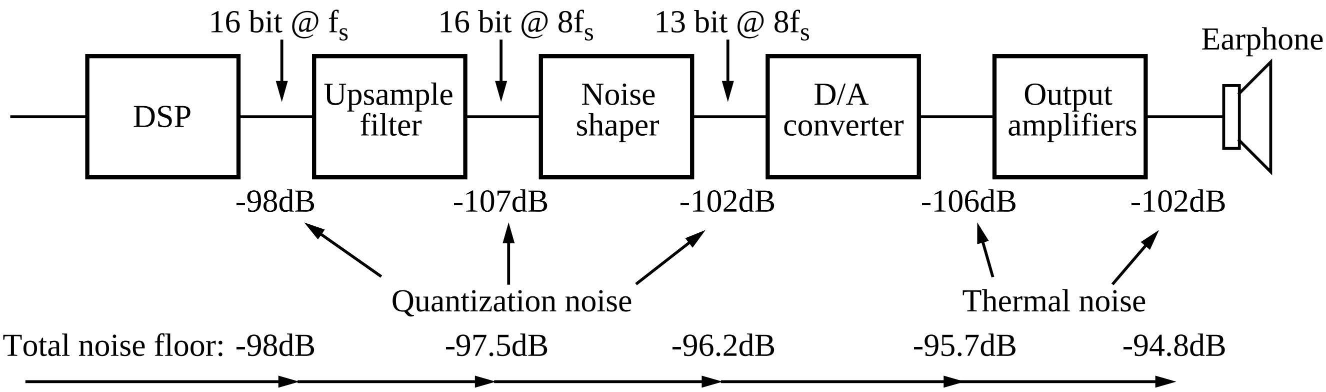


Fig. 17. Analog backend noise budget

Class A/B output stages are driven from the raw battery voltage, while the input stage is fed from the 2.1 V supply. By using the battery voltage also as reference for the D/A converter overload of the amplifiers is prevented. Amplifier bias current is 70 uA, needed to guarantee stable operation.

Digital circuitry.

Fig. 18 shows a block diagram of the most important parts of the digital circuit. The first decimating filter at the input and the last interpolator and noise-shaper at the output are hard-wired, while the low-frequency operations are performed by the DSP core. S/N of the digital path is specified at -93 dB. Cycle budget is 64, but the actual 'algorithm'budget is 46. Because of the optimized architecture this is sufficient for rather complex algorithms.

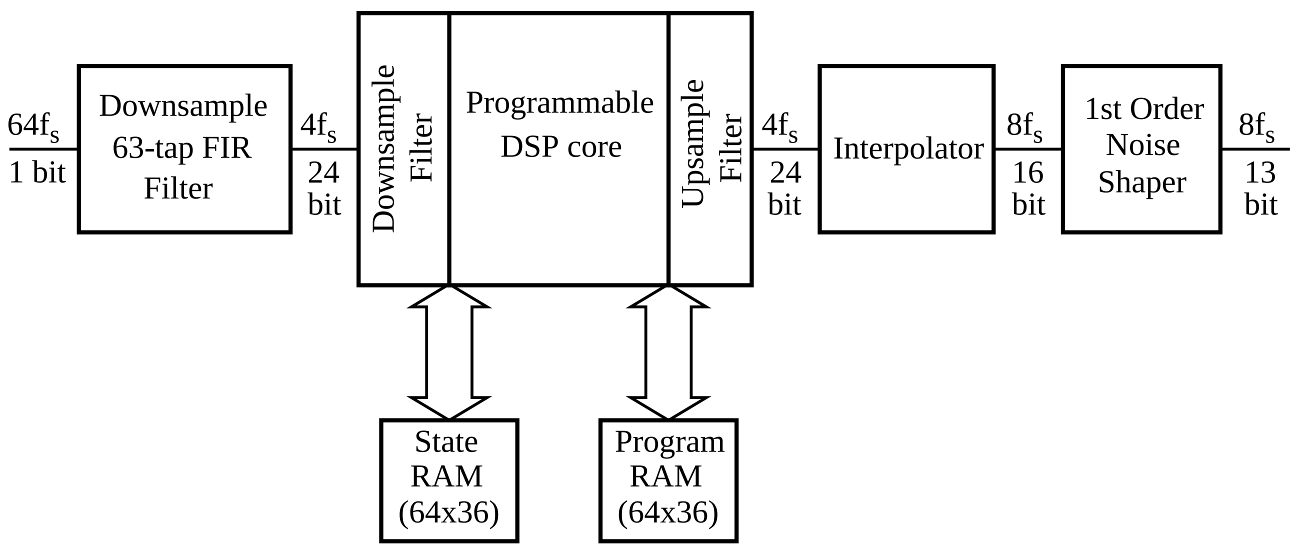


Fig. 18. Block diagram of digital circuit

The DSP architecture is shown in Fig.14. The core is optimised for minimum power consumption by using three application specific datapaths (alu, adaptor, agc) in parallel. The adaptor is optimised for calculating wave digital filter (WDF) structures. It calculates one adaption of a WDF every clockcycle. In a general - purpose processor this would take at least four clock cycles. The agc datapath is optimised for automatic gain control units. The alu datapath allows efficient arithmetic and logical operations. By putting these dedicated datapaths in parallel, the number of clockcycles required for acceptable processing power is limited so an internal clock of 1 MHz is adequate. The hearing aid program is loaded via an infra-red remote control unit. The internal oscillator is frequency-locked to the received signal, so absolute filter accuracy is determined by the remote control oscillator.

IR receiver.

The IR amplifier is an two-stage AC coupled amplifier followed by a comparator as shown in Fig. 19.

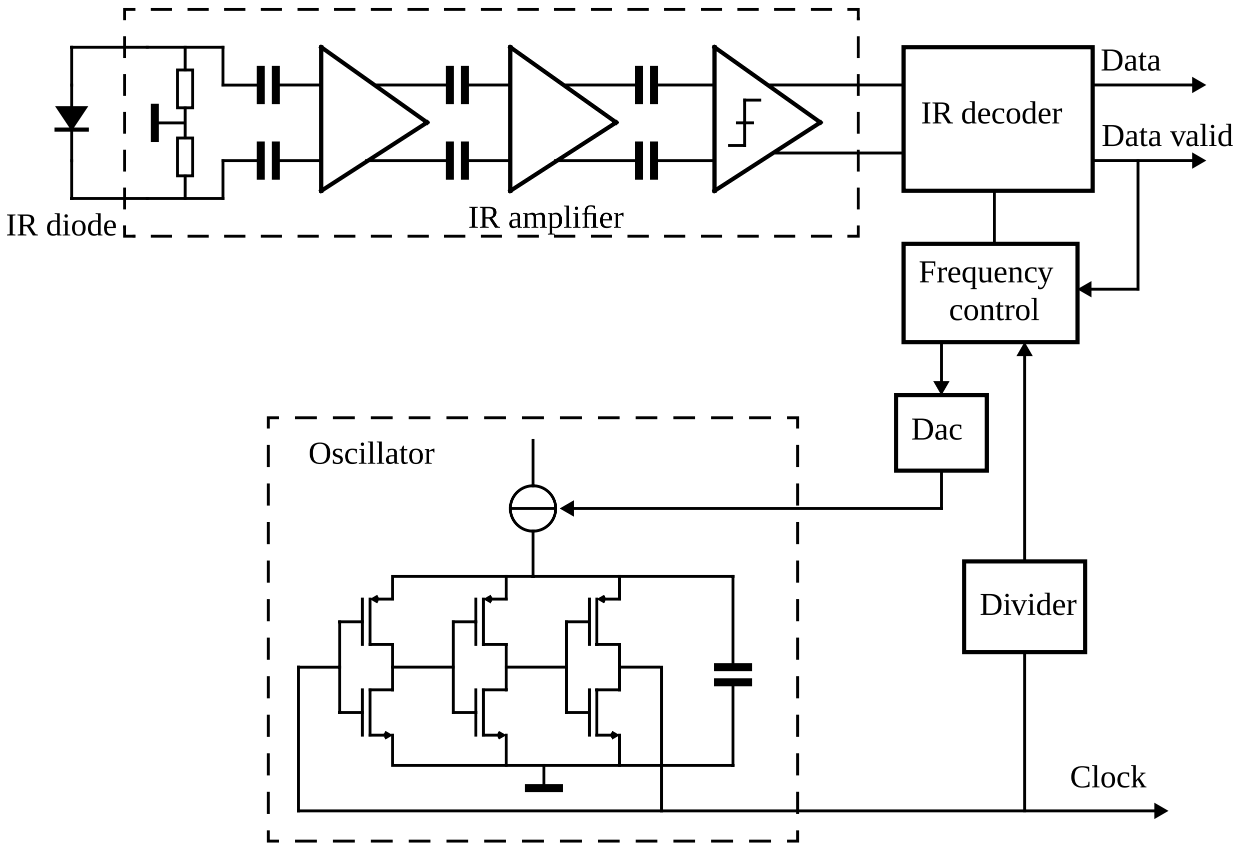


Fig. 19. IR receiver and clock oscillator.

A third order band-pass characteristic for the incoming IR signal, with a modulation of 36 kHz, is achieved by passive high-pass filters used for AC coupling and a low-pass characteristic due to natural bandwidth limitation. The input signal varies from 50 nA to 50 uA, with LF spurious signals up to 100 uA. The gain per stage is limited to prevent overload conditions due to spurious signals. A built-in threshold at the comparator input limits the switching on noise in the absence of transmission, as this would lead to unnecessary power consumption.

As this part is switched on continuously, power consumption had to be less than 10 uA drawn from the raw battery. A resistor across the IR diode prevents it from entering in the forward mode in case of backlight. A digital decoder demodulates the output of the comparator and detects if the incoming data is a valid data packet. The master oscillator generates the clock for the digital circuitry. During reception of a valid data packet, a frequency-lock loop locks the oscillator to the incoming IR data rate of 36 kHz.

Power analysis.

Power in the front and backend is defined by the transducers, independent of IC technology. To start with the tail, the maximum power delivered to the speaker is given by the battery voltage, varying between 0.9 and 1.6 V, and the speaker impedance, about 200 Ohms, giving a maximum undistorted power at nominal battery voltage of about 10 mW. The average output power measured according the IEC norm is about 0.13 mW, taking an average current of 700 uA, when delivered by a class B amplifier.

Though higher output levels up to 100 mW are sometimes required in case of severe perceptual losses, these are not used for 'in -ear' solutions.

At the input side the microphone requires a stabilised voltage of 0.95 V, drawing a current of 20 to 50 uA. The output voltage of the used microphones varies from 12 to 28 mV. This figure related to the required dynamic range of 80 dB over 8 kHz, gives a minimum input noise impedance of 10 Kohm. About 60 uA of the A/D current consumption of 100 uA is needed to bias the input stage. Total current consumption of the chip is 1.5 mA with no input signal of which 700 uA for the analog part, 700 uA for the digital part and 100 uA for additional circuits. Looking at these figures, the average current consuption independent of technology will be some 800 uA and current taken by the system 1400 uA. Analog consumption is not expected to diminish dramaticly, digital consumption for the same complexity certainly will.

Conclusions.

A single chip programmable DSP with all the necessary analog interface circuits for a hearing aid, produced in a conventional 0.8 micron process has shown that both size and power consumption are at present within the range, needed for these applications. Digital processing shows to be attractive with regard to power consumption as soon as high dynamic ranges and moderate signal bandwidths are required. Adding 6 dB dynamic range in a digital filter means just adding one bit in wordlength, increasing power consumption by five procent or less, compared to increasing power four times in an analog solution. For future developments increased complexity can be expected while future power reduction is limited by properties of external sensors.

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