ACOUSTIC ENERGY OUTPUT OF A LOW FREQUENCY ULTRASONIC SURGICAL EQUIPMENT

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Abstract: The aim of this article is to define the necessary acoustic energy input delivered to the operational area during standard procedures with ultrasonic surgical knifes. Acoustic energy, and likewise the acoustic power depend on ultrasonic probe operational frequency, displacement of the probe tip and load of the medium that surrounds the probe tip. In real operational conditions the ultrasonic probe tip acts as a point source of omnidirectional spherical waves; however when the probe tip is operating near a "soft" or "rigid" boundary the sound source should be described as an acoustic dipole.

List of Symbols

E	= acoustic energy, (J)
P	= acoustic power, (W)
I	= acoustic intensity, (W/m^2)
S	= surface area, (m^2)
t	= time, (s)
k	= wave number, (1/m)
λ	= wavelength, (m)
a	= radius of the probe tip, (m)
d	= diameter of the probe tip, (m)
r	= distance between the observation point and acoustic center, (m)
ξ_m	= peak displacement of the probe tip, (μm)
f	= operating frequency, (Hz)
ω	= angular frequency, (1/s)
$\dot{\xi}$	= $\omega\xi$ = vibrating velocity, (m/s)
R_S	= radiation resistance, (Ω)
p	= effective sound pressure, (Pa)
ρc	= specific impedance, (kg/(m^2 s))
$\ddot{\xi}$	= $\omega\dot{\xi}$ = acceleration of the vibrating tip, (m/s^2)
Q_m	= source strength function, (m^3/s)
$R(\theta)$	= directivity pattern of the sound field

Introduction

The ultrasonic surgical knife is an universal and promising surgical equipment, capable of performing different types of minimally invasive surgical interventions and is used in general surgery, ophthalmology and especially in neurosurgery.

This low frequency ultrasonic surgical equipment is essentially a rod type resonant vibrating system, consisting of a half-wavelength ultrasonic transducer made of piezoceramic or magnetostrictive material, and a resonant waveguide in the form of a specially shaped velocity transformer, driven by an electronic generator.

The low frequency ultrasonic surgical equipment is able to:

- deliver sufficient amount of acoustic power into the operational area
- transmit acoustic energy over the saline and induce proper static presssures in the necrosed tissue (intervention area)
- concentrate mechanical energy in the narrow zone near the ultrasonic probe tip

Standard operational procedures of removing cataracteous tissue with conventional ultrasonic phacoemulsification probes, operate at 55 kHz (continuous wave), have peak displacement amplitude of the distal end of the ultrasonic probe of about 100 μm, estimated total acoustic energy of about 3 J and the phacoemulsification time is 120 seconds [1]. Equal procedure, performed with the laser assisted shockwave probe, operating in single or repetitive pulse mode (approx. 600 pulses), needs total estimated acoustic energy of about 400 mJ.

For tissue fragmentation or cutting the tip of the ultrasonic probe must produce necessary amount of acoustic energy, which is defined by

$$E = P\ t = I\ S\,t = \left(\frac{1}{2}\dot{\xi}_m^2 R_S\right)t = 2\pi^2 f^2 \xi_m^2 R_S\, t \qquad (1)$$

Having in mind that the acoustic intensity I is defined as acoustic power P emitted through the unit surface S, there exists an optimal operating frequency of the transducer and it is in range from 25 kHz to 36 kHz [2,3,4,5,6]. Increasing operating frequency towards the range from 35 kHz to 60 kHz would demand a decrease in dimensions of the ultrasonic transducer which would result in a decrease of the vibrating tip peak amplitude ξ_m . In real application conditions for ultrasonic knifes operating near 25 kHz probe tip peak-to-peak displacement amplitudes in the range of ξ_{pp} = 355 μm have been reported, while for ultrasonic knifes operating near 36 kHz this values are much lower, i.e. 210 μm [7].

The value of probe (peak) vibrating amplitude is important because by increasing vibrating amplitude tissue fragmentation and cutting is intensified, and the operational procedure becomes more efficient. Taking into account possible undesirable side effects to the healthy tissue in the vicinity of the operational area, for each individual procedure (phacoemulsification, cutting of tumours tissue, removal of dental plaque, transsection and merging of bone tissue, dissection of carotid artery etc.), optimal vibrating amplitudes, which ensure best final result, must be determined.

Standard ultrasonic knifes for minimally invasive surgical applications, operating near 25 kHz, have probe tip diameter of less than 2 mm. Therefore the ratio of the radiation surface diameter ($d = 2a$) and the wavelength at 25 kHz is $d/\lambda << 1$ and is independent of the probe tip shape (which may be a hemisphere, sphere, cone, or a flat surface). Such wave ratios implicate that the probe tip acts as an ultrasonic point source, emitting symmetrical spherical waves. At an observation point, placed at sufficient distance r from the source, an almost plane wave of acoustic intensity I will exist:

$$I = \frac{p^2}{\rho c} = \rho c \dot{\xi}^2 = \rho c (\omega \xi)^2 \qquad (2)$$

Although there are no conventional devices for direct measurement of acoustic intensity I, expression (2) indicates indirect solution to that. The effective sound pressure p can be measured directly by means of a hydrophone. The probe tip peak displacement ξ_m can be measured accurately by an optical microscope or by the laser vibrometer method, as well as the vibrating velocity $\dot{\xi}$ or acceleration of the probe tip $\ddot{\xi}$ [8]. It is important to emphasize that in the standardized measurement procedure the ultrasonic probe must not be loaded. Its peak vibration displacement will therefore differ from that of the ultrasonic surgical knife operating in real conditions. For this case the acoustic output power P can be measured indirectly by means of the calorimeter method. On the other hand, if the excitation is known, the acoustic output power may be calculated from the impedance characteristics of the unloaded and loaded ultrasonic transducer.

The emitted acoustic energy E (1) depends also on active radiation resistance R_S. This physical quantity characterizes transmission of the acoustic energy from transducer towards the medium, which implicates that R_S has influence on the electroacoustic efficiency coefficient η_{ea} of the transducer. The active radiation resistance can be calculated from the measured electrical input impedance of the ultrasonic probe [9].

Acoustic monopole – omnidirectional point source

Ultrasonic transducers which are used in low frequency ultrasonic devices for operation in acoustic free field conditions emit spherical waves of the monopole type.

In our experimental work, an endoscopic ultrasonic probe with operating frequency of 25 kHz and probe tip diameter $2a = 1{,}6$ mm has been used, with a wavelength/diameter ratio of $\lambda/d \approx 38$ [10,11].

On the basis of the radiation impedance equation [12], and with $\lambda/(\pi d) >> 1$, the radiation resistance is determined from:

$$R_S \approx \rho c S \left(\frac{2\pi a}{\lambda}\right)^2 \approx \pi \rho c \left(\frac{S}{\lambda}\right)^2 \approx \rho c S (ka)^2 \qquad (3)$$

According to (3) radiation resistance depends on specific acoustic impedance, surface area of the acoustic source and the ratio of wavelength to transducer dimension. Due to the self oscillating mass which creates reactive energy between the probe tip and the medium in the near field of the probe, the radiation resistance R_S (real part of the radiation impedance Z_S) is much smaller then the radiation reactance X_S. Namely in this area right beneath the probe tip the vibrating velocity of medium particles $\dot{\xi}$, decreases with the square of the distance from the sound source. In this "nonwave" zone condition $kr < 1$ or $r \leq (\lambda/6)$ is met, which can be used to calculate the zone area. In our case, for the transducer operating at $f = 24{,}5$ kHz, the nonwave zone is a sphere of $r \leq 9{,}7$ mm radius. In this area, where tissue fragmentation and cutting with ultrasonic knife takes place, hydrodynamic effects dominate over wave effects.

In the ultrasonic far field, i.e. where $r >> \lambda$ is satisfied, the radiation resistance R_S is proportional to the surface area S of the sound source and does not depend on the operating frequency. Therefore it can be said that the radiation resistance R_S of spherical waves in the far field corresponds to the radiation resistance of plane waves. In this area the effective sound pressure p is inversely proportional to distance r between the observation point and the sound source origin. According to [12] can it be calculated from

$$p = \frac{1}{2r}\sqrt{\frac{\rho c}{\pi} P} \qquad (4)$$

The equation shows that the emitted acoustic power and energy can be calculated from measured values of the effective sound pressure p [12]. On the other side on the basis of (1), the acoustic power P is determined by

$$P_P = \frac{1}{2}\dot{\xi}_m^2 R_S = \frac{\rho}{8\pi c}(\omega Q_m)^2 \qquad (5)$$

(where index $_P$ denotes a point source)

The quantity $Q_m = S\dot{\xi}_m = 4a^2\pi\dot{\xi}_m$ can be described as the time derivative of the volume enclosed by the sphere of radius $r = a$ and is refferred to as the source strength or volume velocity or even as a productivity of a simple harmonic sound source. According to (5) the acoustic power of any arbitrary shaped sound source of small wave dimensions (including ultrasonic surgical knifes), with known operating frequency f and specific acoustic impedance ρc of the surrounding medium is determined exclusively by the volume velocity Q_m of the sound source [13].

On the basis of (1) and (5), for the point source of spherical waves (acoustic monopole), acoustic energy E for efficient tissue fragmentation and cutting is determined by

$$E = \frac{\rho\,\omega^2 Q_{\mathrm{m}}^2}{8\pi\,c} t \qquad (6)$$

(6) shows the obvious dependence of emitted acoustic energy E on radiation surface area S, operating frequency f, probe tip peak displacement ξ_{m} and volume velocity Q_{m}.

Acoustic dipole – directional sound source

Without regard to the calibration procedure in free-field conditions, or to real operational procedures with ultrasonic surgical knives, two boundary cases are present where the acoustic terms of operation are considerably different. The first one is when the ultrasonic surgical knife operates near a "soft" acoustic boundary (air/water or air/tissue). Namely, the ratio of specific acoustic impedances of air and water (or tissue) is approx. 3600, so 99,94 % of the incident waves to the medium boundary are reflected, i.e. a total reflection of sound waves is present. The "rigid" acoustic boundary has no such explicit acoustic meaning. It is present only when the ultrasonic knife is used near bones. Typical example of such operations are neurosurgical operations in the vicinity of skull bones where the ratio of specific acoustic impedances between the bone and saline (or brain tissue) is approx. 5 [14, 15, 16]. In this case only a part of the acoustic energy is reflected from the boundary, while the other is, due to absorption, transformed into heat. Reflection of sound waves can affect the:

- directivity pattern of the sound field $R(\theta)$, effective sound pressure p in the observation point, vibrating velocity $\dot{\xi}$ of the medium particle, propagation conditions of sound waves,
- characteristics of ultrasonic transducers, radiation resistance R_{S}, and most important, the acoustic power P.

Sound field of a point source of ultrasonic waves, placed at depth h from a "soft" acoustic boundary, correspond to the sound field of an acoustic dipole. The influence of the "soft" acoustic boundary with regard to reflection of sound waves, can be represented by a virtual mirror sound source placed at the height h above the medium boundary (Fig. 1). It is important to emphasize that the mirror sound source must have the same acoustic power P with opposite phase.

The sound field of the described theoretical model of an acoustic dipole near the "soft" acoustic boundary, completely corresponds to the sound field of the an acoustic dipole with opposite phases, placed into free field conditions.

According to [17], for a directional acoustic source (radiator), the emitted acoustic power P_{DS} is given by

$$P_{\mathrm{DS}} = I_{\max} \iint_S R^2(\theta)\,dS = \frac{\rho\,c\,k\,Q_{\mathrm{m}}^2}{8\,\pi}\left[1 - \frac{\sin\frac{4\pi\,h}{\lambda}}{\frac{4\pi\,h}{\lambda}}\right] \qquad (7)$$

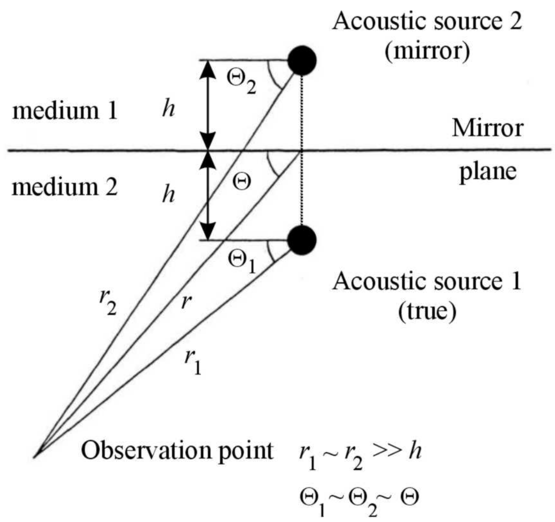

Figure 1: Acoustic dipole modelled by a virtual source

In this case, by comparing (5) and (7), the radiation resistance can be expressed as

$$R_{\mathrm{S}} = \frac{\rho\,c\,k^2\,S^2}{4\,\pi}\left[1 - \frac{\sin\frac{4\pi\,h}{\lambda}}{\frac{4\pi\,h}{\lambda}}\right] = \frac{\rho\,c\,k^2\,S^2}{4\,\pi}\sigma_1 \qquad (8)$$

Comparison of the emitted acoustic power P_P of the point source in free field conditions and the emitted acoustic power P_{DS} of the dipole source ("soft" acoustic boundary), gives

$$\frac{P_{\mathrm{DS}}}{P_{\mathrm{P}}} = 1 - \frac{\sin\frac{4\pi\,h}{\lambda}}{\frac{4\pi\,h}{\lambda}} = \sigma_1 \qquad (9)$$

The diagram of this function is presented on Fig. 2.

In a similar way, the sound field of the theoretical model of an acoustic dipole near the "rigid" acoustic boundary, completely correspond to the sound field of an acoustic dipole with the same phases, placed into free field conditions. The emitted acoustic power P_{DR} of such an acoustic dipole is given by

$$P_{\mathrm{DR}} = I_{\max} \iint_S R^2(\theta)\,dS = \frac{\rho\,c\,k\,Q_{\mathrm{m}}^2}{8\,\pi}\left[1 + \frac{\sin\frac{4\pi\,h}{\lambda}}{\frac{4\pi\,h}{\lambda}}\right] \qquad (10)$$

In this case, by comparing relations (5) and (10), the radiation resistance can be expressed as

$$R_{\mathrm{S}} = \frac{\rho\,c\,k^2\,S^2}{4\,\pi}\left[1 + \frac{\sin\frac{4\pi\,h}{\lambda}}{\frac{4\pi\,h}{\lambda}}\right] = \frac{\rho\,c\,k^2\,S^2}{4\,\pi}\sigma_2 \qquad (11)$$

A comparison of the emitted acoustic power P_P of the point source in free field conditions and the emitted acoustic power P_{DR} of the dipole source ("rigid" acoustic boundary), gives the following relation, also presented on Fig. 2.

$$\frac{P_{DR}}{P_P} = 1 + \frac{\sin\frac{4\pi h}{\lambda}}{\frac{4\pi h}{\lambda}} = \sigma_2 \qquad (12)$$

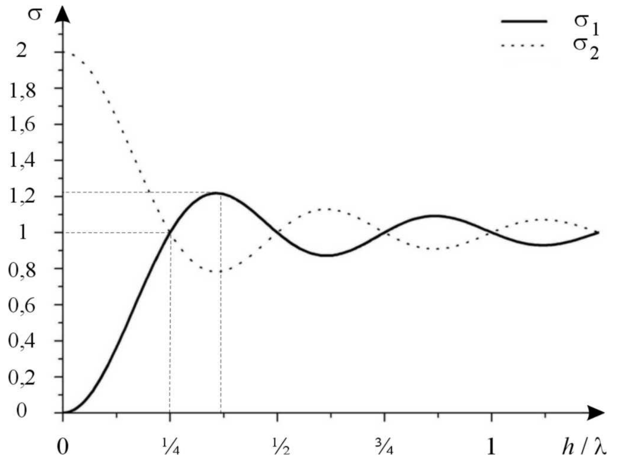

Figure 2: $\frac{P_{DS}}{P_P}$; $\frac{P_{DR}}{P_P}$ as function of depth h

Conclusion

Emitted acoustic power of each ultrasonic transducer with small wave dimensions, does not depend on transducer shape, but on the operating frequency and, in the medium of known specific impedance, only on the source strength function Q_m. There is a significant difference in the emitted acoustic power of the ultrasonic surgical knife operating near "soft" and "rigid" acoustic boundaries, especially in the area where the depth is $h < (\lambda/4)$. Namely, near the "*soft*" acoustic boundary, when $(h/\lambda) \to 0$, the emitted acoustic power P decreases toward its minimum value. But, with the increase of depth, the emitted acoustic power P also increases and at the value of $(h/\lambda) \approx 0{,}375$ reaches its maximum value $P_D \approx 1{,}2\,P_T$. With further increase of depth, i.e. for $h > (\lambda/4)$, the radiation resistance tends toward the radiation resistance of the point source in free field conditions. On the basis of (5), (7) and (10) adequate conclusions on measuring conditions and delivered output acoustic power of the used ultrasonic surgical equipment can be made.

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