ACOUSTIC - INDUCED TURBULENCE IN BUBBLES

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Abstract Air bubble interface dynamics are considered and subjected to acoustic forcing. The bubbles are generated in a tank of water and subjected to variable frequency sound. High-speed video (in excess of 1000 fps) recording of the onset and subsequent surface wave dynamics reveals complex pinching and release of micro-bubbles, both external and internal to the main bubble. The acoustic signature reveals complex frequency interaction; feedback control of these frequencies is proposed using a novel method.

INTRODUCTION

Open cardiac surgical procedures and particularly those using cardiopulmonary bypass inevitably introduce large amounts of air into the circulation. Trapped air circulates through the body until it is either absorbed or lodged in small capillaries in the brain, heart or other organs to result in a myriad of adverse events including death [1]. Air introduced during cardiac surgery correlates highly with neurological dysfunction and this is associated with a reduced quality of life even at 5 years following cardiac bypass surgery [2]. The long-term objective of the current research is to improve methods of detection, quantification and removal of air introduced during open cardiac procedures.

THEORY

The Rayleigh-Plesset (see [3, 4]) equation (1) describes the dynamics of a bubble that undergoes purely spherical motion. The linear harmonic solution to the Rayleigh-Plesset equation gives the natural frequency of spherical motion, the 0^{th} or breathing mode of the bubble, equation 2:

$$R\ddot{R} + \frac{3}{2}\dot{R}^2 = \frac{1}{\rho}\Big[\Big(p_o + \frac{2\sigma}{R_o}\Big)\Big(\frac{R_o}{R}\Big)^{3\gamma} - \frac{4\mu\dot{R}}{R} - \frac{2\sigma}{R} - p_\infty(t)\Big] \tag{1}$$

$$\omega_o = \frac{1}{2\pi R_o}\sqrt{\frac{3\gamma p_o}{\rho} + \frac{6\gamma\sigma}{\rho R_o} - \frac{2\sigma}{\rho R_o} - \frac{4\nu^2}{R_o^2}} \tag{2}$$

where R=R(t) is the bubble wall radius, the single and double dots represents first and second time derivatives, respectively, ρ is the liquid density, p_o is the equilibrium static pressure, σ is the gas-liquid surface tension, R_o is the equilibrium bubble radius, γ is the polytropic index of the gas, μ and υ is the liquid dynamic and kinematic viscosity, respectively, $p_\infty(t)$ is the applied external pressure, and ω_o is the natural frequency in Hz.

When spherical bubble motion oscillations become unstable and spherical symmetry is lost, the analysis of the bubble dynamics becomes exceedingly complex. The first issue is to describe or specify the bubble shape. A solution that has been extensively used [3, 4, 5] for non-spherical bubble dynamics is a spherical harmonic expansion of the bubble radius:

$$r(\theta,\phi,t) = R_o + \sum_{n,m}^{\infty} a_{nm}(t)Y_n^m(\theta,\phi) \tag{3}$$

where $r(\theta,\phi,t)$ is the instantaneous bubble, a_{nm} is the amplitude of the spherical harmonic component of order n and degree m, and $Y_n^m(\theta,\phi)$ is the normalized spherical harmonic shape distortion of the bubble surface.

On linearized theory, the pressure field due to shape oscillations of the bubble surface decays with radial distance (r) like $r^{-(n+1)}$ as shown by Plesset [3]. Hence, these modes have been thought to produce a negligible emission of pressure waves far from the bubble surface. Longuet-Higgins [6] has shown using nonlinear theory the distortion modes produce a monopole radiation of pressure waves that decays as 1/r, the same as the breathing (0^{th}) mode of the bubble. The nonlinear analysis reveals that the frequency is twice the basic frequency of the distortion mode and the sound amplitude is proportional to the square of the distortion amplitude (a_{nm}).

We have developed a multi-frequency acoustic procedure that breaks-up larger air bubbles (1-2 mm diameter) into smaller bubbles (figure 1). Evidence is lacking in terms of precise cut-off size at which point air emboli become clinically significant. It is, however, generally accepted that large emboli are more clinically significant than small ones

[7]. Small potential emboli that remain following application of our break-up technology are less likely to be associated with adverse events than larger bubbles such as those introduced during cardiac surgery.

The objectives of the current series of experiments will be to further investigate the effect of a time dependent, large bandwidth frequency signals on the surface oscillations and instabilities of large bubble dynamics before, during and following break-up. The specific objective is to find the ideal signal to break-up bubbles with maximal effect with minimal time, power and cell damage.

Our experimental rig automatically produces a bubble of known size and then exposes it to a known acoustic signal. The dynamics and break-up are captured at 30,000 fps using a high-speed camera. The bubble fission rate is measured using an edge detection software and feedback to create a new signal. This process is repeated until the maximum bubble fission rate is found.

Our primary goal is to find the combined frequencies for optimal bubble break-up (i.e., maximal effect with minimal time, power and cell damage). The hypothesis is that if we break-up potential air emboli within the arterial system using acoustic methods to a size equal to or smaller than 40 μm, micro-bubbles should be more likely to travel harmlessly through the body until being diffused into the blood. Acoustic signals have been used previously for cavitation both within medicine and industry but these used single frequency ultrasonic transducers [3] which are insufficient for breaking-up large air bubbles (i.e. > 100 mm) like those introduced during cardiac surgery.

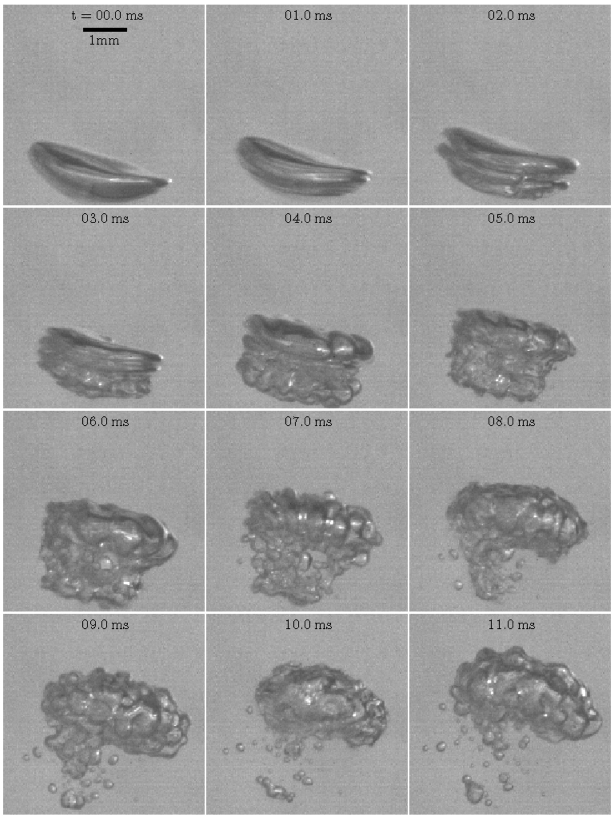

Figure 1. 1.6 mm radius bubble break-up evolution sequence using a linear tone sweep from 1.5 to 1.6 kHz acoustic signal; images shown at 1000 frames per second.

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