Surrogate Lung Material for Impact Studies: Development and Testing

Hamid Khalili Parsa, Aleksandar Karac, Alojz Ivankovic

Abstract This work is focused on the development of a surrogate lung material that not only reproduces the dynamic response of a human lung under various loading conditions but also allows the analysis of the extent and distribution of damage, thus potentially eliminating the practice of live animal testing. The surrogate material consists of polyurethane foam and fluid-filled gelatin microcapsules. In order to characterize surrogate material, various tests were conducted on microcapsules and surrogate material itself. Initially, the bursting pressure of the microcapsules was investigated using low and high rate compression tests. A bursting pressure of around 5 bar was obtained which is comparable to the reported lung overpressure at injury level. Furthermore, low and high rate compression tests were conducted on the surrogate lung specimens to obtain the stress wave speed in the material. The wave speed was found to be well in the range of the reported values for both animal and human lungs (16-70 m/s). A CT scan analysis was carried out before and after the impact tests to study the damage. The damage analysis was then compared to the Bowen curves. Excellent agreement was obtained.

Keywords surrogate lung, compression test, microcapsule, Bowen curve

I. INTRODUCTION

The exposure of the human thorax to high external loads can cause severe injuries to internal organs, particularly to lungs, and can be fatal in many instances. The most critical are rapidly changing loads caused by impacts and blasts, resulting from car crashes, collisions and explosions. The most common lung injuries are edema and hemorrhage. These injuries impair the oxygen transport mechanism and cause secondary injuries to the brain. This is due to the fact that the lungs are more susceptible to damage than the other organs in the human thoracic region. Besson and Saegesser [1] reported 15 injuries per million per day with 11% relating to road accidents. According to the World Report on Road Traffic Injury Prevention 2004, 1.2 million people are killed in road crashes every year [2] while in 2009 there were over 5 million police report crashes in the United States, resulting in over 2 million injuries and 33,808 fatalities [3].

In order to better understand the human dynamic response to various impact loads and also to determine the injury risk to various body regions as they are exposed to severe forces developed in the impact, physical and computational models are used in blast and impact studies. This understanding is considered essential to develop the next generation of personal protective equipment, vehicles and structures to protect against the primary blast injury (PBI). The current commonly used physical models consist of a combination of biological organs such as lungs, heart, livers, etc. from live animals and human cadavers as well as synthetic representations of other parts of the body. While using live animals in various tests, such as body armor trials, allows the effects of blunt trauma to be investigated accurately, it does raise concern among animal welfare groups and poses moral, ethical and legal problems for both governmental and non-governmental organisations engaged in such activity. The development of a successful physical and numerical lung model will allow an unlimited number of tests to be carried out under a variety of conditions. This would therefore provide a better understanding of blast impact and predicting PBI in human lungs as well as eliminating the practice of live animal testing or human cadaver testing (where used) and can be incorporated into an artificial recreation of the human torso for accurate testing of realistic situations.

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There are numerous studies related to this type of problems available in literature. The Naval Research Laboratory developed a measurement device (GelMan) designed specifically to study the dynamic response of the human thorax to a variety of loading conditions [4]. Roberts et al. [5] and Merkle et al. [6] developed a human surrogate torso model with internal thoracic organs. This surrogate used solid bio-simulant materials for all internal organs, including the lungs (silicon foam). Saraf et al. [7] studied the dynamic response of the human tissues including lungs from cadavers using the Kolsky bar technique. Post-mortem human subjects were also used to establish force-deflection response corridors for blunt torso ballistic impacts to the torso [8]. It was demonstrated with cadaver specimens that the viscous criterion [9] can be used as a measure of injury risk for ballistic impacts [10]-[11]. The tissue injury mechanisms occurring during the thoracic response to rapid chest wall displacement have been differentiated into high-frequency injuries caused by wave propagation and low frequency damage as a result of displacement-related causation [12]. Injuries caused by both mechanisms can occur in ballistic impacts. Gryth et al. [13]-[15] and Cannon [16] studied behind armour blunt trauma resulting from the impact of projectiles on personal armors. Gryth et al. [13]-[14] tested human-sized swine and reported severe behind armor injuries when shot with 7.62 mm bullets, and recently extended their study to look at pathophysiological changes and compensatory mechanisms that occur early after severe lung contusion [15]. Nevertheless, many similar studies were conducted by the military, in particular to test the integrity of body armour protection suits, and the results are not readily available.

In addition, various authors have been investigated the damaging effects of the passage of waves through the human lungs [17]-[21]. As the wave propagates it induces stresses and strains within the alveolar wall. It has been proposed that trauma is caused by either a combination of spallation, implosion, inertial or pressure differential effects in the lung micro structure [17],[22]-[24]. It was hypothesised that the main reason for the injuries is excessive tensile stress in the alveolar walls caused by the re-expansion of the alveoli which collapsed during the propagation of the compressive pressure wave. This is a popular theory adopted by many in this field [25]-[27].

To the authors' knowledge, there has not been any documented attempt to create a surrogate lung model that will simulate not only the behaviour (i.e. wave propagation) but also the associated damage to the lungs as a result of impact loading. Instead, tests are conducted on live animals (mice, rabbits, sheep and pigs) and these results are suitably scaled to predict blast injuries in humans [28]-[29].

Our aim is to develop a surrogate lung model that not only reproduces the dynamic response of a real lung under a variety of conditions but also allows the extent and distribution of damage to the lungs to be studied. This would eliminate the need for the practice of live animals and human cadavers for testing purposes. To achieve this aim, the surrogate lung material should fulfil two main criteria: (i) its stress wave speed should be similar to that of a real lung of 16-70 m/s [30]-[34], (ii) the extent and distribution of damage resulting from impact at appropriate pressures should be similar to the lung overpressure at injury level, i.e. above 5 bar [20],[35]-[36].

II. MATERIALS AND METHODS

In this study, the surrogate lung material was designed as a mixture of polyurethane foam and fluid-filled gelatin microcapsules. A closed-cell polyurethane foam with a theoretical wave speed of 40 m/s was used to mimic the spongy material of the lungs, which draws support from Grimal et al. [20], where they state that during the passage of a stress wave, the air is trapped in the alveoli for frequencies above a few hundred hertz which are typical for high rate impact loading. The closed-cell structure of the foam satisfies this.

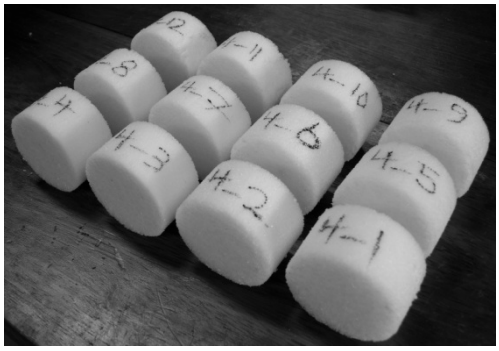

Fig. 1. Cylindrically shaped surrogate lung specimens.

The polymeric foam comes in two liquid parts, which are to be mixed in proper mass ratio. Prior to mixing microcapsules were added. The components were mixed for 10-15 seconds using a steel T-bar attached to a hand drill. The mixture was finally poured into a waste PVC pipe (of inner diameter of 51 mm) which was already lubricated with PTFE/Silicon oil spray and then placed vertically. The pipe and the mixture were left to set for 10 minutes. The pipe, now filled with foam, was then cut into 30 mm thick slices using a band saw. From this, cylindrically shaped surrogate lung specimens (measuring 51 mm in diameter and 30 mm in height) were manufactured (Fig. 1).

Gelatin microcapsules filled with Barium Sulfate solution are used to simulate the damage of alveoli in blast tests in terms of hemorrhage and bursting pressure. Barium Sulfate solution is used as it is detectable by X-rays and therefore can be easily scanned and used for analysis of damage. In other words, the microcapsules are designed to break at a certain pressure, the critical pressure, representing the "injury criterion" for the model. These microcapsules are compressed in low and high rate experiments, being in the stress state similar to that of alveoli in the lung during impact.

In order to fully characterize surrogate lung material, both the microcapsules and surrogate material must be separately analysed. For this, a number of compression tests were carried out on each of them to ascertain their mechanical properties.

It has to be noted that only typical representatives of each test are given here. More details can be found in [37].

Characterization of Fluid-Filled Microcapsules

In order to characterize the microcapsules, a series of compression tests were carried out on individual microcapsules at low and relatively high test speeds. For low rate tests, each microcapsule was compressed using the Hounsfield compression/tensile machine between two flat steel surfaces, as shown in Fig. 2-left, at a constant rate of 200 μm/min until bursting occurred. The load was measured using a 10N load cell. The profile of the deforming microcapsule was also recorded during test by a Nikon digital camera attached to a travelling microscope.

By measuring the contact area (A) between the microcapsule and the plates and using the recorded load at burst (F_b), critical or bursting pressure (p) can be calculated as:

$$p = \frac{F_b}{A}, \quad (1)$$

Relatively high rate compression tests were also carried out on a stationary microcapsule at four impact speeds of 0.44, 0.82, 1.2 and 3 m/s using a small drop-weight apparatus specially designed for this purpose (Fig. 2-right). The loading rate was varied by varying the drop height of the striker. A Kistler piezoelectric dynamic load cell was used to record the load while the deformation of the microcapsule was recorded by a Phantom high speed camera.

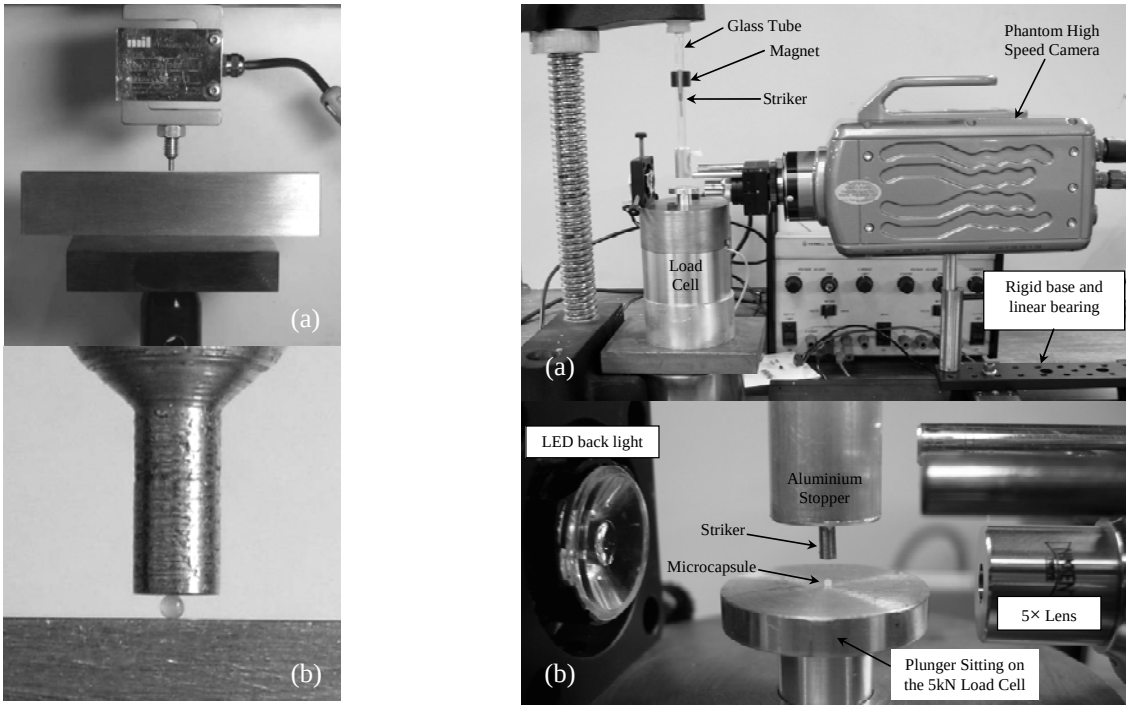

Fig. 2. Compression test on a single microcapsule for low (left) and high (right) rate tests (a) experimental set-up (b) magnified view.

Characterization of Surrogate Lung Specimens

Surrogate lung specimens were loaded in uniaxial compression tests under a variety of loading rates ranging from 1 to 4.82 m/s, producing strain rates in the range 30-160 s^{-1}, and various compression ratios (10-90%). Here, compression ratio refers to the ratio of the deformed distance over the original thickness, i.e. the engineering strain in compression. The tests were performed using an instrumented drop-weight impact-testing machine as shown in Fig. 3 with load being measured using a Kistler load cell. Hence, stress-strain in compression and stress-time histories were obtained.

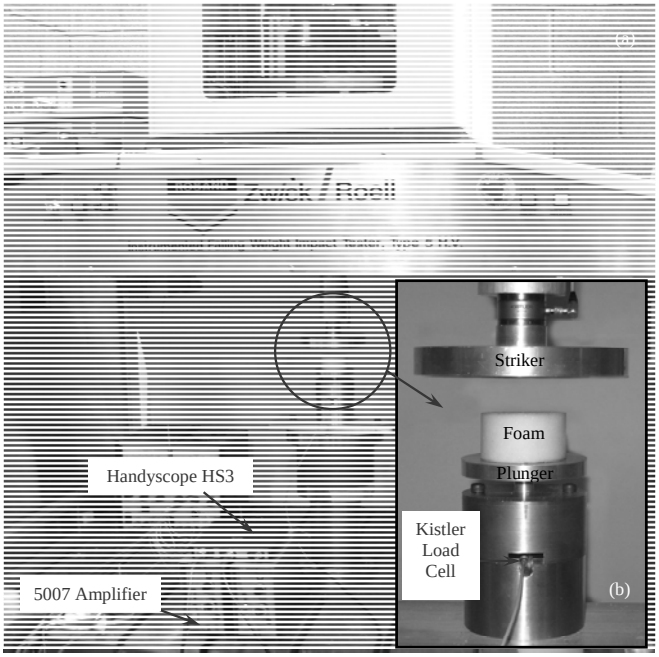

Fig. 3. High rate compression test on a surrogate lung specimen: (a) experimental set-up (b) magnified view.

Stress-strain histories were used to estimate the modulus of elasticity (*E*). Fig. 4 shows a typical engineering stress-strain curve for a test at 2.5 m/s and 80% compression. *E* is obtained by fitting a straight line to the initial

elastic region of the stress-strain curve (up to 0.1 of engineering strain). The slope of the fitted line, as indicated in Fig. 4b, represents the Young's modulus (E=88.74 kPa). By having the value of E and the density ρ of each specimen, the stress wave speed (C) was obtained from:

$$C = \sqrt{\frac{E}{\rho}}, \quad (2)$$

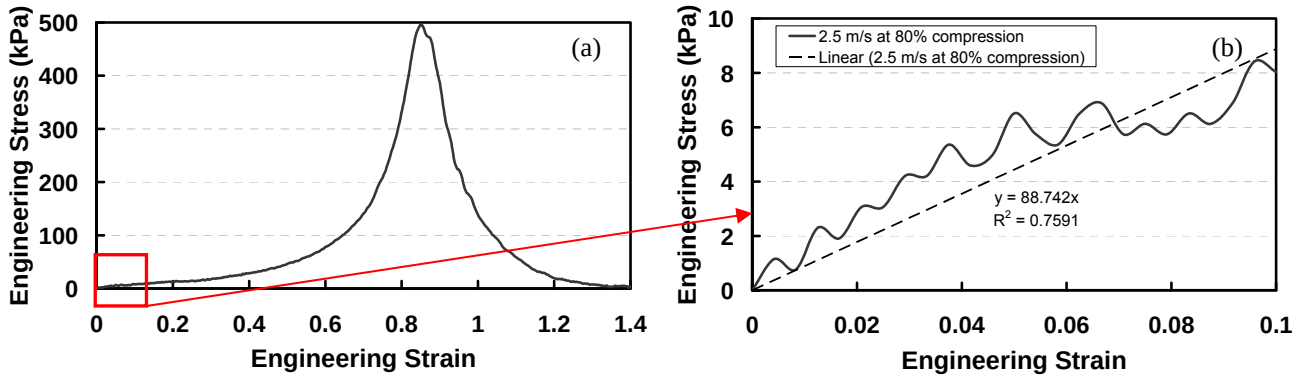

Fig. 4. Stress-strain curve at 2.5 m/s: (a) full history (b) calculation of E.

On the other hand, stress-time curves were used for comparison with Bowen curves [28]-[29], where each point indicates a combination of the applied equivalent stress/pressure and the duration for which that stress is applied, used to predict the lethality of free-field blast waves. To this end, an experimental stress-time curve was approximated into a corresponding rectangular stress pulse, having the same area as the area under the actual stress-time curve, but using different representative stress, and hence stress pulse duration. Three approaches based on different definitions of representative stress are proposed. In approach 1 (Ap1), the representative stress, σ_{15}, is taken to be equal to 1.5MPa (corresponding to a hydrostatic pressure of 5 bar, known as the threshold pressure for lung injury [20],[35]-[36].For approach 2 (Ap2) and 3 (Ap3), the representative stress was equal to the maximum stress (σ_{max}) and the average stress, σ_{avg}, (average between σ_{15} and σ_{max}) from the stress-time history, respectively. Calculated stress/pressure level with accompanying pulse duration are then plotted on Bowen diagrams.

Furthermore, in order to characterize the surrogate lung specimens in terms of damage initiation and distribution within the specimens, they were scanned individually using the CT scanner located in St. Vincent's University Hospital in Dublin. In these scans, a slice thickness of 1mm and exposure of 120kV were employed before and after the impact tests to inspect the damage as represented by burst microcapsules.

III. RESULTS

In terms of manufactured quality, the developed surrogate lung specimens were found to be excellent with minimum amount of large holes and gaps. Also, a uniform distribution of microcapsules was achieved in all cross-sections of specimens as indicated by CT scans.

For microcapsule tests, a bursting pressure of 5.4±0.68 bar was obtained for low rate compression tests. In terms of relatively high rate experiments, the bursting pressure decreased when the impact speed increased as shown in Fig. 5. A maximum pressure of 6.8±0.58 bar was obtained at 0.44 m/s and thereafter the pressure reduced to a value of 2.58±0.7 bar at 3 m/s, with minor decrease from 1.2 m/s.

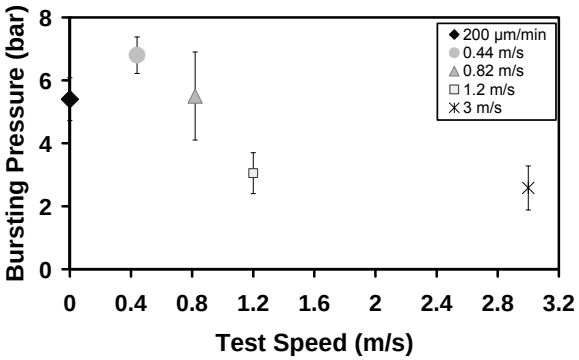

Fig. 5. Comparison of bursting pressures at various impact test speeds

In terms of surrogate lung material characterization, modulus of elasticity and density are measured and the stress wave speed was calculated using equation (2). The value obtained was in the range of 20-30 m/s.

The virgin and the tested surrogate lung specimens are examined before and after impact by a CT scanner to analyze the extent and distribution of damage. Fig. 6a illustrates the CT scan image of a virgin surrogate lung specimen, while Fig. 6b shows the same slice of the same surrogate lung specimen after being impacted at speed of 4.82 m/s and compressed up to 85% of the original thickness of 30 mm. The bright and shiny white spots in Fig. 6 are the microcapsules, the gray region is the polyurethane foam and the black spots are tiny air holes.

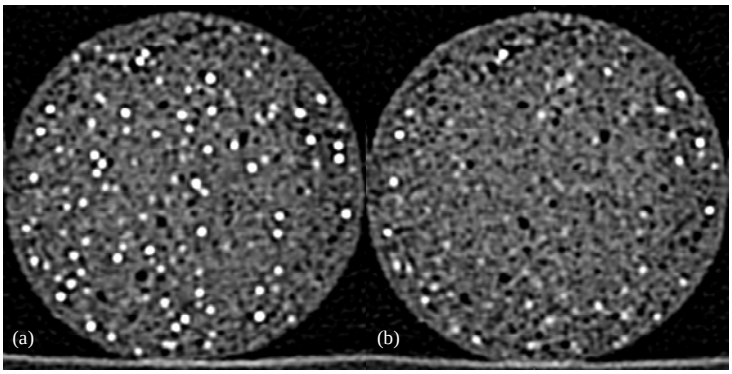

Fig. 6. CT scan images of a surrogate lung specimen (a) virgin, (b) damaged. Radius of cross-sections is 51 mm.

In each slice of the virgin surrogate lung specimens, the average number of microcapsules was approximately 61. This was assigned as the reference number of microcapsules in any undamaged plane and compared with the planes of the impacted specimens by counting the remaining undamaged microcapsules after impact (i.e. bright and shiny white spots). The ratio of damaged to undamaged microcapsules was used to define the percentage of damage occurring in the specimen. This was then plotted as a function of location from the anterior to the posterior of the impact site. Fig. 7 shows the statistical distribution of burst microcapsules through the same specimen shown in Fig. 6 after damage together with the data on distribution of microcapsules before the test. The calculated mean damage was 78%.

It is interesting to note that there are still some unbroken microcapsules present after the impact, mainly at the periphery, as illustrated in Fig. 6b. The presence of these unbroken microcapsules is due to the fact that the highest hydrostatic stresses are concentrated in the middle of the specimen in the compression experiments. If the specimen is in uniaxial compression, then the hydrostatic pressure ($\sigma_m = \sigma_1/3$) is uniform across the thickness. However, the surrogate lung material is not homogeneous and the local micro-structure might cause differences in pressure across the section (non-uniform bursting of microcapsules across the section).

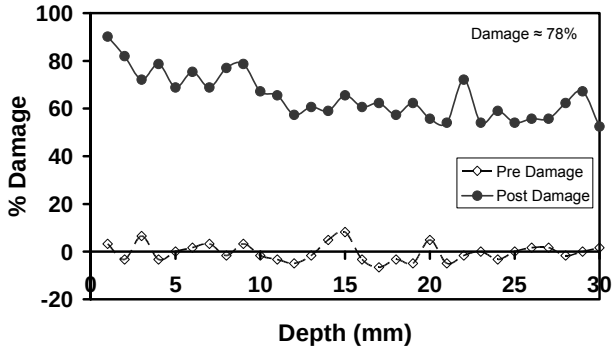

Fig. 7. Distribution of burst microcapsules before and after damage at 4.82 m/s and 85% compression

The procedure was then employed to all tested specimens and results are compared with Bowen curves as discussed in Section II. Figure 8 shows a typical stress-time graph for a test at 2.5 m/s and 88.33% compression. The area under the curve is approximated and the duration of the overpressure for each approach (the width of each rectangle) is then obtained. Ap1 and Ap3 were only applicable to those tests where the pressure exceeded the threshold lung injury of 5 bar.

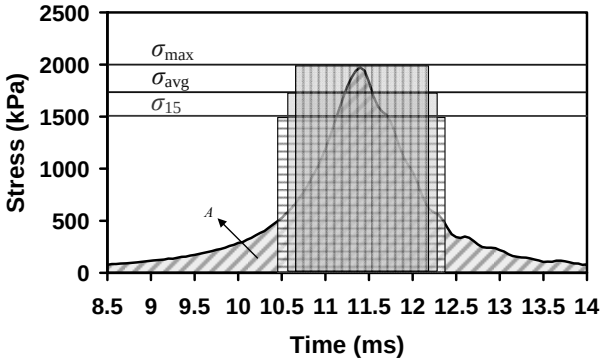

Fig. 8. Stress-time graph with three approximations

IV. Discussion

Based on the experimental data on microcapsules, a bursting pressure of around 5 bar was measured for low loading rate of 200 μm/min which is in excellent agreement with the reported lung overpressure at injury level of above 5 bar. Higher bursting pressures in comparison to the threshold for lung injury were calculated at higher speeds of 0.44 and 0.82 m/s (up to 6.8 bar) while the pressures decreased below the threshold to 2.58 bar at 3 m/s.

Results from high rate compression tests on surrogate lung specimens indicated an increase in stress wave speed with increase in test speed, due to an increase in modulus of elasticity. The stress wave speeds of 20-30 m/s were calculated for tests performed at 1-4.82 m/s which are well within the physiological range of typical lungs of 16-70 m/s [30]-[34]. In particular, these wave speeds are close to the values reported for calf lungs of 24-30 m/s [30], horse lungs of 25-70 m/s [31] and that of human lungs of around 30 m/s [32].

The results obtained from CT scans of impacted surrogate lung specimens showed that for the same compression ratio, a higher test speed gives a higher percentage of damage, as indicated in Figure 9.

The three approaches used for comparison of experimental results on surrogate lung material against the Bowen curves showed limitation in use of Ap1 and Ap3. On the other hand, Ap2, where the maximum stress values were used as representative stress for damage analysis, had no limitations and can be used for all cases. Corresponding results are given in Fig. 10 for various speeds and compression ratios and compared to the percentage of survival and the threshold lung damage in the Bowen curve. The results of CT scan analysis of the damage distribution in the surrogate lung specimens are shown in the figure legend; “D” stands for Damage and “C” for Compression Ratio.

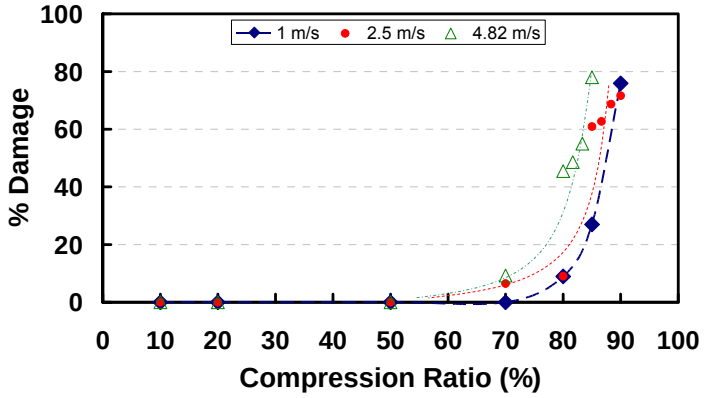

Fig. 9. Damage analysis for various test speeds and compression ratios

It was shown that the experiments at speeds of 2.5 and 4.82 m/s with compression ratios of 90 and 85%, respectively, exceeded the 1% survival limit on the Bowen curves while a survival rate of 5% was predicted for tests at 2.5 m/s and 88.33% compression ratio. Tests at 1 m/s and a compression ratio of 90% were in the range of around 10% survival on the Bowen curves and the CT scan results for this testing condition revealed 75.9% damage while the calculated hydrostatic pressures (3.3 bar) were lower than the limit of 5 bar based on the literature. Survival rates of 30% and 50% were obtained for surrogate lung specimens tested at 4.82 m/s with a compression ratio of 83.33% and 2.5 m/s with a compression ratio of 86.66%, respectively. CT scans showed 55% and 62.7% damage for these two test conditions. One percent mortality (99% survival) was predicted for 4.82 m/s and a compression ratio of 81.66%. CT analysis presented damage of 48.6% for this particular test. For the 2.5 m/s tests at 85% compression survival rate was 90%. The analysis of burst microcapsules using the CT scan images for this case confirmed 60.9% of damage.

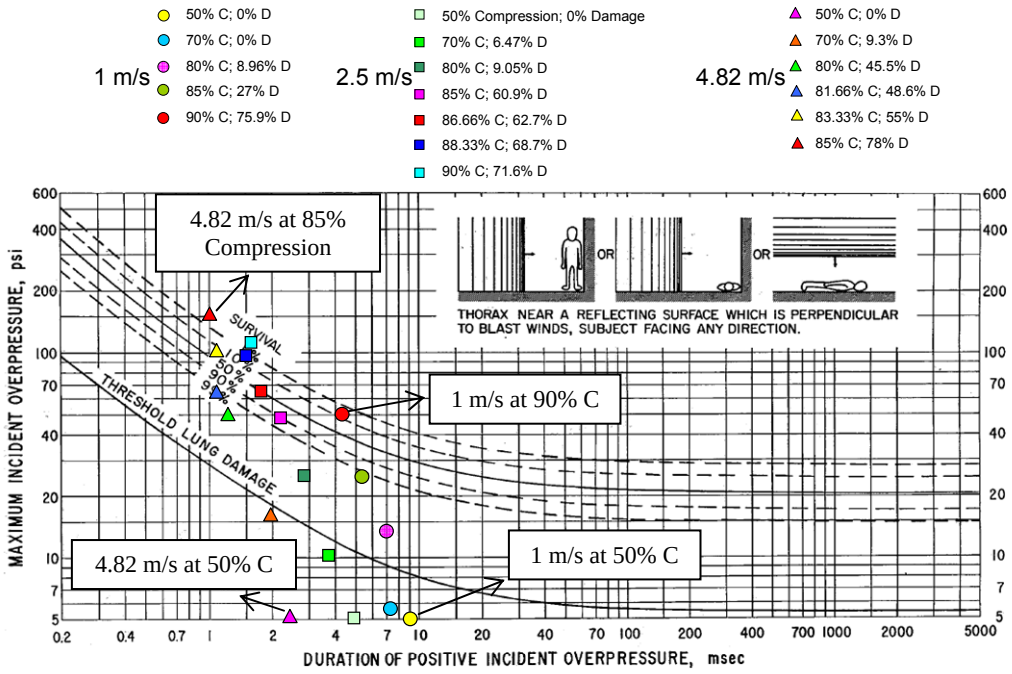

Fig. 10. Damage prediction of experimental results using Ap2

V. Conclusions

Manufactured surrogate lung material fulfils the two main criteria for testing purposes: a similar stress wave speed to that of real lungs and a comparable bursting pressure to that of alveoli at injury level. Therefore, this surrogate material could be used as a replacement for cadaver human lungs as well as the lungs from live animals.

The selected microcapsules may be utilised in the surrogate lung specimens for simulating the damage experimentally. CT scan results clearly demonstrated the magnitude and distribution of damage within the specimen as represented by burst microcapsules. It may be concluded that these results can be used as a quantitative measure of the damage by comparing the state of the microcapsules before and after testing.

The CT scan results, combined with stresses calculated during the compression tests, compared well to the Bowen curves, so the method could be used as a predictive tool for studying the damage to human and animal lungs in the future.

VI. Acknowledgement

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