Mechanical Testing of Artificial Vessels and Tissues for Photoplethysmography Phantoms

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***Abstract*— Various studies have looked at the efficiency of artificial vessel and tissue networks in the study of photoplethysmography (PPG) in an effort to better understand the origin of various morphological features present in the signal. Whilst there are all reasonable attempts made to replicate geometrical features such as vessel depth, vessel wall thickness and diameter etc., not many studies have attempted to replicate the mechanical properties such as vessel elasticity and tissue compressibility. This study reports two methods for tissue mechanical testing for the analysis of vessel elasticity and tissue compressibility. A two-part polydimethylsiloxane (PDMS) was used as a base material for both tissue and vessel construction, and the properties altered by changing the curing component ratio. Tissue compression properties were investigated using an industrially calibrated materials testing device using the protocol from the ASTM 0575-91 testing method. Vessel elasticity was investigated using a custom method and apparatus to report vessel diameter and length change simultaneously. Tissue compressive properties proved reasonably easy to replicate through catalyst alteration, however the vessel elasticity properties were found to be higher than expected at all reasonable catalyst ratios. The property of hyper-elasticity was observed in the artificial vessels though, leading to the conclusion that alternative material recipes or construction methods may be needed to correctly replicate the expected mechanical characteristics.**

***Clinical Relevance*— The latest generation of health monitoring devices, especially those that are wearable and used widely by individuals wishing to monitor their health daily are becoming smarter and more sophisticated in their functionality. The majority of such devices use photoplethysmography (PPG) as their primary monitoring technique. Being able to replicate the PPG in a phantom allows the continued study and development of devices, and to improve their functionality without the continued need for extensive user-testing.**

I. Introduction

Photoplethysmography (PPG) is an optical technique first established in 1937 [1], and it is used to measure various subcutaneous hemodynamic properties with well-established clinical monitoring applications [2]. The PPG waveform derives from the fact that arteries and arterioles within tissue contain more blood during systole than during diastole, and thus more photons are detected by the sensor [3]. The introduction of the pulse oximeter in 1974 advanced the use of PPG-dependent technology significantly as it was adopted into anesthetic practice [4].

The resurgence of PPG technology in the last few decades has been due to the growing demand for personal wearable health technologies, for example smart watches or sport-bands that enable the non-invasive assessment of cardiovascular physiology and potential disease diagnosis [5]. Despite the long-established history of PPG, research into the many causal factors remains a popular area of research [6].

In vitro PPG investigations are an attractive area of study for the investigation of PPG morphology as they address the limitation of *in silico* studies, where the pulsatile component is ignored [7], and *in vivo* studies where it is often impractical or very difficult to induce certain physiological parameters, for example when researching the use of PPG for the non-invasive measurement of blood pressure [8, 9].

In addition, the *in vitro* studies can reduce not only the cost of clinical trials that has been escalating every year [10] but also potential risks of clinical trials significantly [11].

There is a large interest in developing a system that accurately characterizes the true biology of tissue in a more controlled and rigorous fashion. A method to achieve this is using advanced phantom technology. The development of diagnostic systems for applications in healthcare, such as monitoring blood oxygenation, glucose concentration, *in vivo* lactose measurement and several other biological markers through absorption spectra analysis, have all required the use of tissue-simulating phantoms [12-14].

Skin and vessel phantoms have been widely used as test models for a variety of peripheral tissue imaging techniques. These often comprise of a skin mimicking material surrounding a vessel through which blood mimicking fluid can be pumped or perfused. These phantoms can serve many purposes. A main feature would be to use it as a constant reference to compare existing systems against each other. They could also serve functions such as initial systems testing and optimization of signal-to-noise ratios (SNR) during their development stage. The literature on materials used in phantoms is broad [15] and throughout the years several custom multi-modal phantoms have been described.

To design a phantom for PPG research, the optical, mechanical and geometrical properties all must be representative of the desired tissue to be investigated. There are limited optical phantoms available commercially that

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could be used in PPG based technologies to allow in-depth investigation of light tissue interaction.

Recently a novel vessel tissue phantom was developed by this research group for the purpose of PPG morphology investigations [16], and successfully deployed to investigate the effects of PPG sensor contact pressure on PPG signal quality *in vitro* [17]. The method used a polydimethylsiloxane (PDMS) material as a base material for both the vessels and tissues, but only a simple shore-hardness test was conducted to see whether the softness of the material was comparable to that of human tissue, and the elasticity of the constructed vessels had not yet been established.

In this paper we discuss and test techniques to establish the correct tissue compressibility and vessel elasticity according to established known values in the literature.

II. Methods

This section describes the methods used to characterize the mechanical properties of both the custom fabricated PDMS vessels, and the PDMS tissue segments of the phantom. The elastic properties of the vessel under inflation are investigated alongside the compressive strength of segments of PDMS that mimic the tissue portion of a phantom.

A. Custom Vessel Inflation Experiment

The fabrication method of the PDMS vessels [16] can be customized to produce vessels at various elasticities by just altering the catalyst ratio of the PDMS mixture. The manufacturer guidelines for the selected material (Sylgard 184, Dow Inc, US) are to mix the base with 10% of its weight with the catalyst. Through experimentation, the lowest stable catalyst ratio for vessel production was found to be 3.5%, beyond which the cured solution was too fragile to complete the fabrication process, and instead formed as a gel.

Following the methods outlined [16], custom PDMS vessels were produced at 10%, 5%, and 3.5% catalyst ratios. The vessels had a wall thickness of 0.2 mm ±0.04 mm and an outer diameter of 3.07 mm ±0.12 mm which is similar to human radial arteries [18]. The vessels were then cut to lengths between 5 mm and 10 mm and connected to a syringe pump in a closed system show in figure 1. The pump was used to increase the fluid levels in the vessels and cause inflation. The internal pressure (mmHg) was measured by a pressure transducer, and the resulting change in length and diameter was recorded by a digital microscope. A precision ruler was used as a calibration tool during analysis of the captured images to measure dimensions.

For each iteration of the test an initial image was captured with the calibration ruler in focus. The pixel distance between the marks were obtained and then used in the conversion factor for pixels to millimeters. The syringe pump was used to increase the internal pressure from 0 mmHg to the point of failure in 20 mmHg increments. At each step, an image of the vessel was captured with the focus adjusted to sharpen the sides of the vessel. The length was measured from a reference point on the right side of the vessel across the length of the vessel to the black tape. The mean diameter was measured from three points perpendicular to the length line. Figure 2A and 2B show the length and diameter measurements for the 10% catalyst ratio vessel at 0 mmHg and 1000 mmHg respectively.

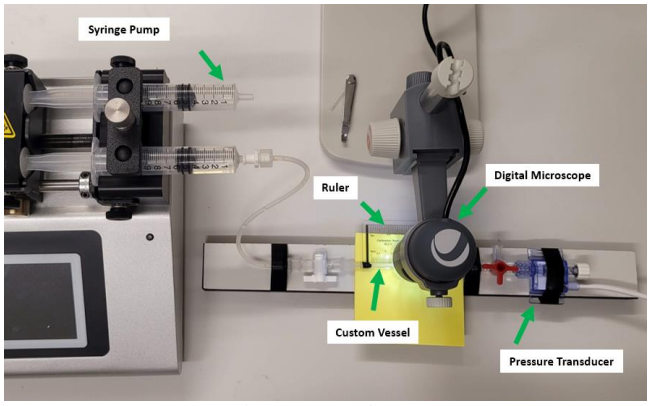

Figure 1. Photograph of the vessel inflation set-up. The syringe pump provided the manual control of internal pressure, with user relayed feedback from the pressure transducer. The digital microscope recorded the parameter changes during inflation by the pump.

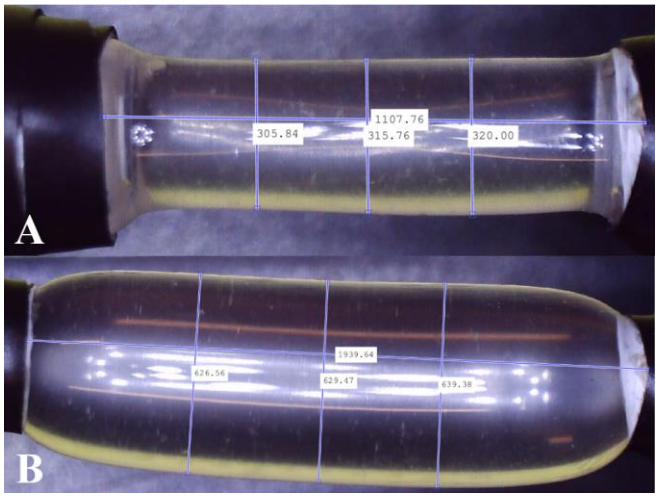

Figure 2. (A). 10% catalyst vessel at 0 mmHg with annotated pixel measurements of diameter and length. (B) the same 10% vessel at 1000 mmHg internal pressure with annotated pixel measurements of diameter and length.

B. Compression Tests on PDMS Tissue

A basic pulsatile phantom consists of a vessel across which pulsatile flow travels through. This vessel is often embedded into a material which mimics the tissue that would surround the chosen vessel in biology. The manufacturers guideline of 10% catalyst ratio will produce a silicone sample with a shore hardness of 43A which converts to a 2.71 MPa Young's Modulus (YM). Through experimentation it was found that by reducing this ratio down to 2%, the elastic properties of the silicone rubber can be manipulated, beyond this point the cured solution resembles a silicone gel.

To aid in the design process of custom phantom fabrication, a dataset for the compressive strength of the PDMS at various catalyst ratio solutions needed to be produced. Eight solutions ranging from 2% to 11% catalyst ratios were poured into 0.5-inch cylindrical molds to cure at room temperature. After a 48hr cure time, the silicone was removed from their molds and cut into 1-inch specimens. These specimens then underwent compression testing (figure 3) in an Instron 5944

in accordance with the ASTM D0575-91 standard for rubber properties in compression.

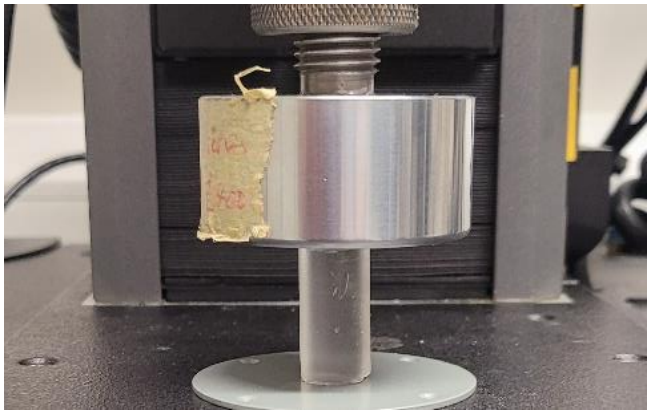

Figure 3. A cylindrical (0.5 x 1 inch) silicone sample under compression testing.

III. Results

A comprehensive set of test data was produced for a range of PDMS formulations for both the tissue and vessel components of the phantom separately. This data was analyzed and converted to the forms seen in comparable *ex vivo* studies for comparison [19-21].

A. Vessel Elasticity Analysis

The elastic response of the 10% catalyst ratio PDMS vessel was found to be too stiff, as can be seen in figure 4 (blue line). Reducing the catalyst ratio does increase the elasticity of the vessel significantly and emphasize the hyperelastic nature that is expected from them, however, even at 3.5% the response is still not in the target region required and referenced in [21].

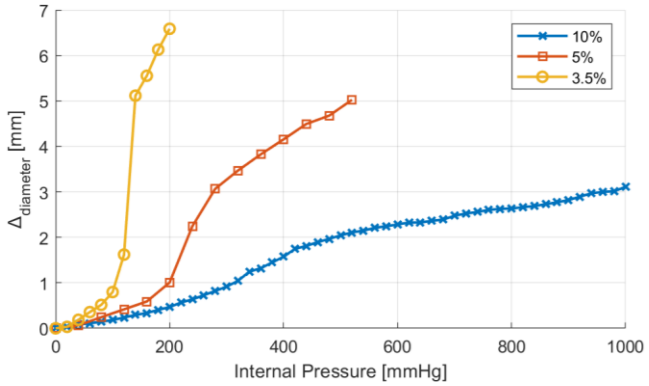

Figure 4. Internal pressure of artery against the measured external diameter change for three catalyst ratios.

B. Tissue Compressibility Analysis

It is observed that as the catalyst ratio decreases so does the YM, as expected. The graph in figure 5 shows the measured YM at the various catalyst ratios. The nearest YM values for tissue from the finger and arm can be found in [19] and [20] respectively, ranging between 0.07 and 0.2 MPa for the finger and 1.03 MPa for the forearm.

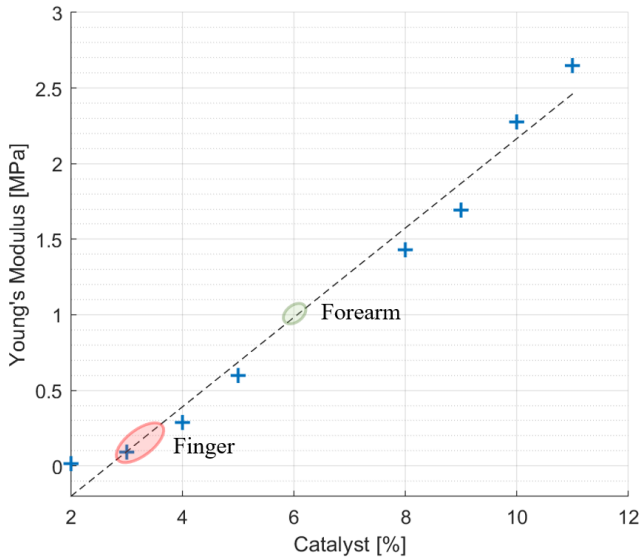

Figure 5. Measured Young's Modulus at different catalyst ratios with the marked YM range for the finger and forearm as seen in the literature [19, 20].

IV. Discussion

The YM decreases as the catalyst ratio decrease. This is expected as the material resembles a soft rubber at 11% and a soft gel at 2%.

The current work in vessel manufacturing is highly encouraging, although it is clear from experimentation and the results shown that the limit of what can be achieved with the current vessel construction has been reached. Significant efforts need to be made to replicate the elastic properties of human vessels closer.

To facilitate further artificial vessel construction, it is recommended that the following work needs to be carried out:

1. Curate a new set of data for the elastic properties of human arteries, the referenced work by Langewouters [21] is unclear in some of the values reported, and the methods used to arrive at the values are only partially given.
2. Explore new elastic materials for vessel manufacture, including different PDMS formulations, latex and polyester. Additives to the materials may also be needed, for example an artificial fibrin, elastin, or collagen substitutes.
3. Explore the manufacturing method used for vessel construction. It is regarded that the current technique, whilst excellent at producing elastic tubes of a thin wall thickness with consistency, cannot accommodate a layering of different types of materials that may be needed to replicate the multi-layer nature of arteries, where each layer has different elastic properties that each contribute differently to the overall elastic nature of arteries.

Of course, the manufacture of human-like tissue and blood vessels are not limited to only *in vitro* PPG investigation studies. The knowledge can be also applied to tissue-engineered vascular grafts [22] and soft tactile robot sensor [23].

V. Conclusion

A novel vessel-testing protocol utilizing digital microscopy and a precision syringe driver was successful developed,

demonstrating accuracy at measuring small incremental diameter changes in artificial vessels. The ASTM 0575-91 testing method was found to be suitable for the artificial tissue compressive measurements.

It has been demonstrated that current manufacturing techniques for the artificial vessels, whilst sophisticated to produce the needed vessel dimensions needs some alteration to the materials used to produce vessels with similar mechanical properties.

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