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Research Article

Exploratory study about apparent elastic properties of human breast

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Abstract: Since the behaviors of human breasts are rubberlike, current research explores elastic properties of human breasts. The elastic behavior of breasts is observed through force-displacement measurements in a laboratory setting and characterized through tensile and shear moduli considering individual breast geometries. The breast geometry was analyzed based on specific dimensions, such as cross-sectional areas and lengths in different orientations. The average tensile and shear moduli are estimated to be 5.04 and 0.96 kN/m^2, respectively, while both vary from individual to individual in wide ranges. The research approach is highly exploratory and examines apparent elastic properties, but the findings make it possible to quantify elastic capability of live human breasts as a whole and compare them with other soft materials. The conclusion suggests that the shear modulus might be more appropriate to characterize the elasticity of human breasts in terms of developing sportswear.

***Keywords*:** breast elasticity, tensile/compression modulus, shear modulus

1. Introduction

Overlying pectoral muscles on the ribcage, female breasts are one of the few external organs that are not supported by skeletal structure. Made of glandular and adipose tissues, there are no muscles in the breast either [1, 2, 3]. Flexible connective tissues, called ligaments, provide a certain degree of support to the breast, and hold the breast tissue in place creating its unique shape [4, 5]. The breast does move along with the thorax, but as a function of the ligaments, together with the breast skin and fat [3], it also presents elastic behavior of its own, bouncing up and down.

In a breast in a vertical motion, three different stages are involved depending on the relative displacement of the breast from the chest: namely, natural, neutral, and elevated or drooped positions [6]. When left intact without any intervention, the breast stays in a state of equilibrium naturally heading downwards because of gravity. The breast mass determines how much it is lowered, and it will also be affected by the properties of the skin encompassing the breast. In case a certain amount of upward force is present by any means, such as a brassiere, the breast is lifted. It enters the neutral condition as the natural drooping is revoked at some point. If the breast keeps moving up beyond the neutral level, it will start being compressed. Specifically, the bottom tissues become stretched, while the top tissues are contracted. On the other hand, when there is a force applied in the opposite

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direction pressing the breast down, contraction happens at the bottom tissues and stretch takes place at the top tissues.

There have been huge aesthetic concerns associated with severely drooped breasts because its shape and movement are considered unattractive to see [7]. Large-breasted women suffer more from this issue since the breast mass is associated [8]. The performance of ligaments might also matter for the large-breasted women. Playing a role to hold and shape the breasts, the ligaments are known to function less as intense displacements of breasts are repeated over time [9, 10].

Apparent elastic behaviors of breasts are also emerging concerns in breast dynamics. The breast undergoes a substantial amount of deformation together with displacement when external force is introduced. This seems to be the major cause of breast discomfort associated with physical exercise. It is believed that the displacement and deformation could be restrained with the help of external support, such as wearing a sports bra [1, 10, 11]. There have been vigorous research endeavors in the past decades to understand breast kinetics and provide the external support for female athletes and exercisers. However, it still remains elusive to calibrate apparent elasticity of breasts and engineer the right amount of support.

Several previous researchers took the approach to investigate breast kinetics by observing breast bounces [1, 11, 12, 13]. The experimental settings had human subjects walk or run on a treadmill to monitor the oscillating motions of their breast induced by exercise. This approach is technically invaluable and can provide scientific background behind breast movements. However, due to some experimental conditions that are difficult to control [10], breast kinetics are extremely complex to analyze.

For example, the exercise-induced forces were inconsistently recurring at random frequency governed by each subject's own strides and gaits [10, 14, 15]. The bouncing force was difficult to quantify because the breast mass is inaccurate and elusive to estimate. In addition, there is also a dynamic damping force to consider in recurring breast bounces that leads to the decay of the amplitude of vibrations [6, 16]. It originates from energy dissipation by diverse types of frictions from between internal molecules to sliding motions. Accordingly, with multiple parameters difficult to take account of, most attempts ended up providing incidental descriptions of breast kinetics and were limited in acquiring meaningful quantification comparable to other elastomers [17].

According to McGhee & Steele [18], the total amount of breast motion is decided by two factors, breast displacement and the number of breast bounces. During treadmill running, breasts showed periodic motions having a complex sinusoidal waveform in three-dimensional directions. Vertical displacement of the breast was reported to range between 4.2 and 9.9 cm. The displacement in other directions was significantly less, 3.0 to 5.9 cm along the sagittal axis and 1.8 to 6.2 cm along the frontal axis. The amount of displacement is large enough to cause sensible discomfort and pain.

Elastic materials consist of polymeric chains tied together with a high degree of flexibility and mobility. When external force is introduced, the chains rearrange their configurations, presenting high deformability. A typical rubber-like material may be stretched up to about 10 times its original length. If the external force is removed, it immediately recovers its original dimensions up to a certain range of stretch. In case of breast stretching or compressing, biological tissues undergo collagen-fiber reorientation [19], presenting good elasticity. In general, high elongation, low elastic modulus, and high recovery represent an excellent degree of elasticity.

Rubbery materials are also incompressible, which means they change their shape rather than changing their volume. Compression in one direction results in extension in the other directions accordingly, and brings about a bulge [20]. The incompressibility is represented by the Poisson's ratio close to ~0.5 [3, 21]. Considering women experiencing aesthetic issues related to breast bulging, it is reasonable to assume that human breasts resemble elastic materials to a certain extent [22].

One of the most basic approaches to characterize the elastic behavior of elastomers is to investigate the relationship between stress and strain. Under the low strain, the ratio of stress and strain is known to be constant and expressed by initial elastic modulus. This value is the same when they are compressed or extended [20]. Stress-strain curves have been a traditional approach widely adopted to characterize tensile behaviors of textile materials. However, it needs to be noted that elastic modulus is strain-dependent, and the stress-strain relationship becomes no longer linear as the strain increases. Also, higher strain leads to hysteresis creating a difference in elastic modulus between compression and extension [20].

Tensile or compression modulus (E) is acquired from a stress-strain curve while a specimen is under the tensile loads pulling the specimen outwards to elongate. Shear modulus (G) is associated with the shear force acting in a direction parallel to a planar cross section of the specimen. Shear deformation takes place in unaligned directions having one end of the specimen change in a specific direction and another end in the opposite direction. Those moduli are defined as follows:

$$E = \frac{\sigma}{\varepsilon} = \frac{Fl_0}{A\Delta l} \tag{1}$$

$$G = \frac{F l_0}{A \Delta x} \tag{2}$$

where σ is stress, ε is strain, F is the force which acts, A is the area on which the force acts, l_0 is initial length, Δl is longitudinal displacement, and Δx is transverse displacement [19, 23].

A good number of studies have investigated the tensile modulus of various human tissues, mostly as diagnostic tools such as the detection of cancerous regions. However, a limited amount of literature was found regarding shear modulus. The levels of tensile and shear moduli of breast and other relevant materials, such adipose tissues and skin, were obtained from these studies and summarized in Table 1 since each must contribute to the apparent elastic properties of breasts. For additional comparison, some synthetic rubbery materials are listed as well.

Table 1. Tensile and shear modulus of soft tissues and materials as found from the literature

	Type of material	Tensile modulus (N/m^2)	Shear modulus (N/m^2)
Comley & Fleck [24]	Porcine adipose	0.4K	–
Alkhouli *et al.* [25]	Abdominal adipose tissue	1.6K	–
Samani & Plewes [26]	Breast tissue	3.6K	–
Sarvazyan *et al.* [23]	Breast adipose tissue	5K–50K	–
Krouskop *et al.* [27]	Breast adipose tissue	18K–22K	–
Krouskop *et al.* [27]	Breast glandular tissue	28K–35K	–
Krouskop *et al.* [27]	Breast fibrous tissue	96K–116K	–
Markidou, Shih, & Shih [28]	Versaflex CL2000X rubber	98K	31K
Silver, Freeman, & DeVore [29]	Thoracic skin	100K	
Sutradhar & Miller [30]	Breast skin	195K–480K	
An, Luo, & Shen [31]	Adipose tissue	270K	–
Sun *et al.* [32]	Adipose tissue	–	4K
Comley & Fleck [24]	Porcine dermis	400K	–
Ramezani & Ripin [21]	Elastomers	400K–700K	–
Lamers *et al.* [33]	Abdominal skin	–	2–6K
Geerligs *et al.* [34]	Adipose tissue	–	7.5K
Zhang *et al.* [3]	Breast skin	0.2–3.0M	–
Zhang *et al.* [3]	Pectoralis muscle	0.08–0.10M	–
Ramezani & Ripin [21]	Nylon plastic	2.1G–3.4G	–

Exploring apparent elastic properties of human breasts, the current research focused on the mechanical characterization of elastic properties that affect breast aesthetics as well as kinetics. Force-displacement measurements were administered in a laboratory setting with live human breasts. To facilitate accurate and precise experimental observations with human subjects, several technical equipment was incorporated, such as 3D body scanner, motion capturing cameras, force gauge, and motorized test stand. Considering individual breast dimensions and geometry, apparent elastic modulus of breasts associated with compressive and shear deformations was characterized. Breast geometry was analyzed from 3D body scan files, where specific dimensional details were approximated, such as cross-sectional areas and diameters in different directions. The approach was highly exploratory since an unconventional experimental setting was implemented and theoretical models of elasticity were applied to explain empirical observations from human subjects. However, this new empirical approach made it possible to quantify the apparent elastic capability of human breasts and compare them with other soft materials.

2. Materials and methods

Ten female subjects were recruited among university students. Their ages ranged from 19 to 25 with the average of 20.4. For clear and obvious observations of breast displacement, the breast size was regulated to be larger than 34B. The average breast volume was estimated to be 603 cm^3. Experimental procedures were approved by the Institutional Research Board (IRB), and the participation was voluntary while monetary compensations were offered.

2.1 *Breast Dynamics*

The experimental setting is illustrated in **Figure 1**. A sling was crafted out of a fine fishing line and a medical tape to elevate breasts during the experiment. The fishing line was tied to create a 55cm-long hoop at the participant side and the opposite side was attached to a digital force gauge (ZTS-11, Imada, Japan). A seat was prepared on the sling (**Figure 1**) by attaching a medical tape to the bottom of the hoop, and this helped the breast sit on the sling more securely. With a series of kinetic pulleys arranged at the ceiling, the string was pulled at the controlled speed of 300 mm/min by a motorized test stand (MH-275, Imada, Japan), where the force gauge was mounted. In this configuration, the force was recorded while lifting the breast.

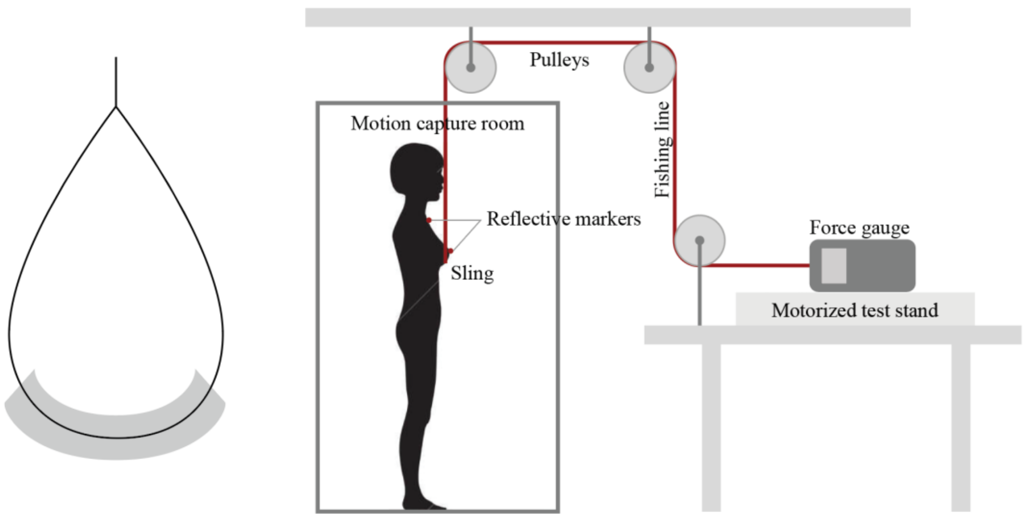

Figure 1. Breast sling with swing seat (left) and experimental measurement setup (right)

Reflective markers were attached to the suprasternal notch and bust point of the participant (**Figure 1**). As the breast moved up, the elevation of bust point was monitored by Vantage motion capture cameras (Vicon Motion Systems Ltd., UK). The vertical displacement of bust point was calculated as the relative location from the suprasternal notch. Not consisting of homogeneous materials, breasts might present anisotropic elastic properties. However, elastic behaviors in one direction, along the vertical (longitudinal) axis, were observed in this research, which matters most in breast kinetics [10]. Although the sling supported the breast from the bottom in the experimental setup, it was assumed that the force was applied to the entire breast and the displacement of the breast was represented by the bust point. The lift was precisely administered and monitored by the test stand, force gauge, and motion cameras.

Three different breast states were hypothesized, as illustrated in **Figure 2**, during the breast lifting experiments depending on the relative position of the breast to the chest: namely, natural, neutral, and elevated positions [6]. The natural state was when the breast stayed in physical equilibrium free from any external force or support. In this state, the breast naturally headed downwards because of gravity. As the string was pulled, the breast was elevated, and the sag was revoked. That moment was expected to be a neutral state. As the breast kept going up beyond the neutral level, it started being compressed and deformed. Specifically, the bottom tissues became stretched, while the upper tissues contracted. The degrees of elevation and relevant breast geometries are also illustrated in **Figure 2**, which will be explained in detail later.

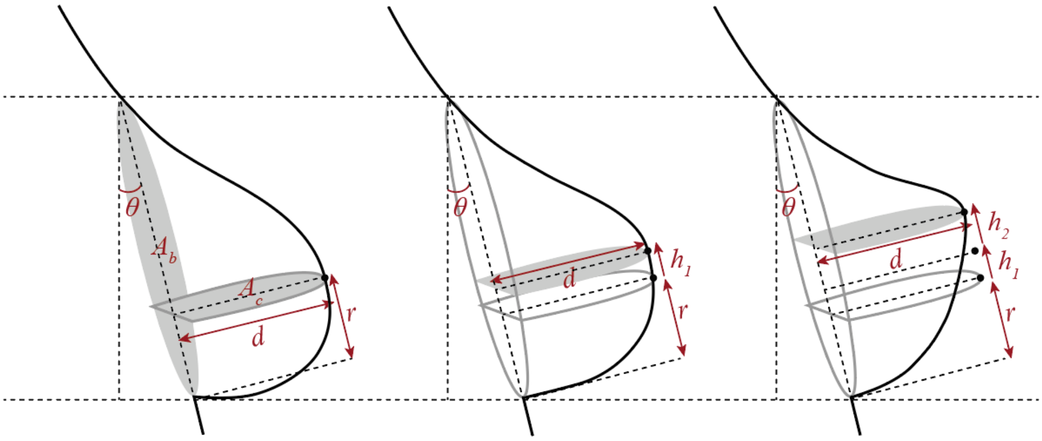

Figure 2. Three states of breast position: natural (left), neutral (middle), elevated (right)

2.2 *Stress-Strain Data Analysis*

Figure 3 shows the scatter chart of force and displacement data collected from one of the participants. Based on the degrees of breast elevation, the breast was expected to be lifted (h_1) from the drooped position to the neutral state during the first portion of the curve (h_1- F_1), which did not always look linear (α). The oscillating trends of raw data were considered to indicate the sling slipped from the breast skin several times during this lifting motion (**Figure 3**). As the breast passed through the neutral point after the breast sag was taken back, it entered the phase of deformation and started being suppressed upward (h_2) because there was no more room to move up and it. Associated with the second portion of the curve (h_2- F_2), apparent elastic behavior could take place. A visible slope change (β) was identified. The sling seems to be positioned tight to the breast surface and the slippage did not take place as much as before. Unlike the prior section, a clear linear trend line was easily observed in the second portion of the curve. Additional raw datasets are present in the result section for visual demonstrations of data variability.

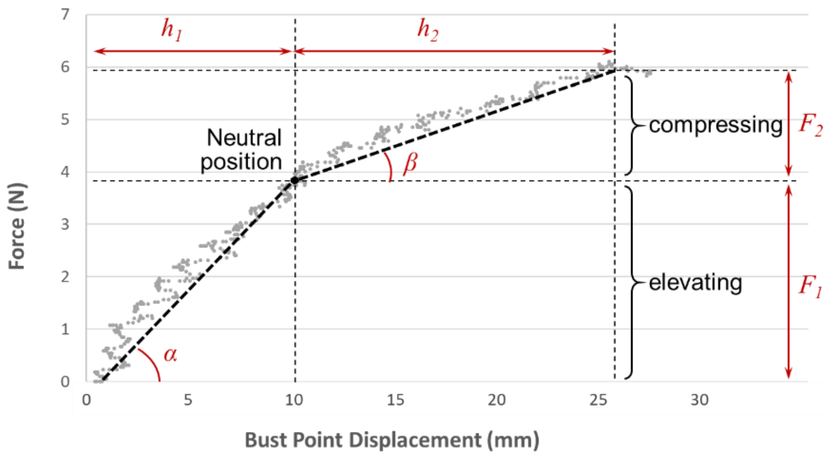

Figure 3. Example dataset of force and displacement during the experiment

2.3 *Breast Geometry*

Each participant was body-scanned to acquire anthropometric data of their breast. It was done by a 3D body scanner (SS20, Size Stream, Cary, NC) while their upper underwear was off. Breast-relevant anthropometric data included bust girth and underbust girth, on which the calculations of conventional bra size have been based.

Although this method is not scientific nor systematic to characterize breast dimensions, this provides a good insight to understand breast size in general.

The scan file was imported to 3D image analysis software (GeoMagic Design X, 3D Systems, Rock Hill, SC) for additional breast geometry analysis. Through this, it was possible to study breast-specific dimensions and skeletal structures of the ribcage more in depth. That included breast radius (r), breast depth (d), breast base area (A_b), cross-sectional area (A_c), breast base angle (θ), and breast volume (V) (**Figure 2**). Breast geometry matters a lot when characterizing its apparent elastic properties since the apparent elastic behavior depends on its dimension. For example, human thorax is skewed vertically against the normal direction of gravity and therefore, the breast bases are located along the tilted direction (**Figure 2**). The angles of breast base (θ) are different from participant to participant depending on their body shapes and postures. Once the breast base angle is measured for each participant, it enabled the researchers to take geometric approaches and calculate the directional force and displacement.

The breast geometry relied on the breast base, and the procedures to identify the breast base were as follows. The bust point (BP) was located on the most prominent position of the breast surface (**Figure 4**). By observing the transversely cross-sectional body contour at the BP level, medial and lateral points of the breast were created. In a similar manner, the inferior point of breast was identified after observing the cross-sectional body contour along the plane initiated from BP and perpendicular to the line connecting the medial and lateral points. A referential plane for breast base was generated connecting the medial, lateral, and inferior points, which ended up invading the armpit and upper arm areas (**Figure 4**). This referential plane was tiled every 0.5° anchored on the inferior point until obtaining a completely closed loop. The breast base planes identified in this way took a water drop shape as shown as a blue line in **Figure 4**.

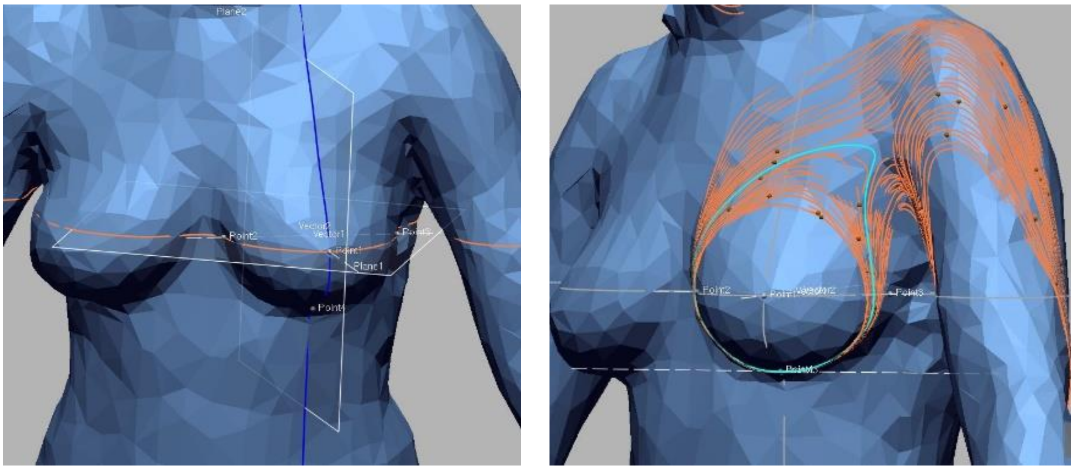

Figure 4. Procedures to identify breast points (left) and breast base (right) in 3D space

After the breast base was identified in this way, the breast geometries were characterized for each participant in terms of breast radius (r), breast depth (d), breast base area (A_b), cross-sectional area (A_c), breast base angle (θ), and breast volume (V) as shown in **Figure 2**. Since the force and displacement was measured along the gravity direction, those measures were processed with the breast base angle to acquire the directional force and displacement along the ribcage. It needs to be noted that the breast depth and cross-sectional area might have been affected by the breast positions during the elevation and compression, but that was not taken into consideration in this research.

3. Results and Discussions

The experimental data was collected for force and displacement as well as breast dimensions. As described in the research method, the point of neutral breast position was identified after finding the inflection point of the curve. Two additional raw datasets are present in

for visual demonstrations of data similarity and variability. Fluctuations of data points, which indicated sling slippage, also helped to confirm where the inflection point was located. Due to the variability of individual datasets, finding the point of neutral position was achieved manually by the researcher relying on visual assessment of the slope in each scatterplot. Each dataset looked distinct in terms of the degrees of breast elevation before being compressed as well as the severity of sling slippage, but the inflection point of the curves was able to locate at a certain point as shown in **Figure 5**.

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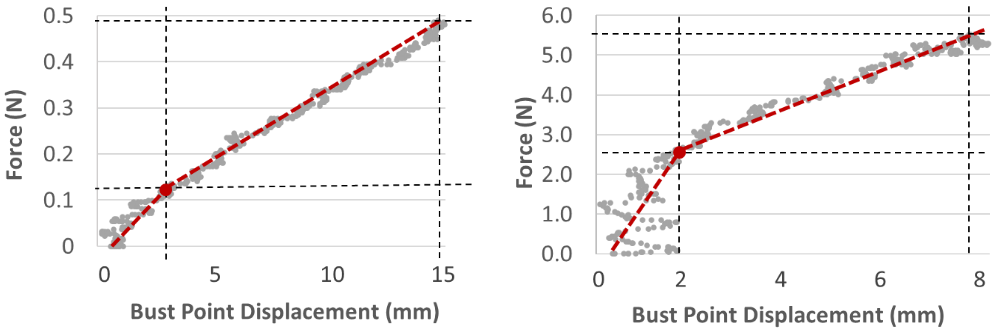

Figure 5. Two additional raw datasets of stress-strain measurements

3.1 *Breast Kinetics*

Experimental measurements from each participant are summarized in **Table 2**. Interrupted by sling slippages that were sometimes more severe than others, the force and displacement measurements varied much from participant to participant during the elevation (**Figure 3** and **Figure 5**). On average, however, it took 1.91 N to elevate the breasts by 2.64 mm towards the neutral point. Larger breasts seemed to experience more elevation (h_1) than smaller breasts. The correlation coefficient between breast volume and elevation distance was 0.7997, indicating a strong positive correlation. This could be explained by severe sagging that took place with large breasts due to their high total breast mass [8]. This led to more elevating force (F_1) associated with large breasts accordingly and the correlation coefficient between the force and breast volume was 0.7201.

Although it was challenging to define clear trendlines for $\tan\alpha$ during the elevation, the average force of 1.01 N was required to elevate the breasts for a unit length (**Table 2**). This had a weak negative relationship with breast volume measurements, showing the coefficient of -0.4845 (**Figure 6**). Being weak, the negative correlation implies that large breasts have been elevated by less amount of force per unit length than small breasts, which may be against a general assumption. It should be noted that human breasts are not homogeneous. The composition of breasts is known to vary depending on their size and large breasts contain relatively more fat than glandular and fibrous tissues [2, 3, 5]. This finding indicates that apparent unit mass of large breasts could be significantly lower than small breasts.

Table 2. Force and displacement data calculated from 10 subjects

	bra size	volume	F_1	h_1	$\tan\alpha$	F_2	h_2	$\tan\beta$
	–	cm^3	N	mm	–	N	mm	–
P01	32DD	1,185	2.89	4.35	0.71	3.24	19.52	0.17
P02	34C	559	2.73	1.50	1.84	3.28	6.20	0.53
P03	30D	296	1.48	1.10	1.55	4.89	8.08	0.62
P04	30E	520	1.19	1.58	0.75	2.93	9.08	0.33
P05	30DD	367	0.72	1.07	0.67	1.79	4.15	0.43
P06	38C	530	0.83	1.18	0.76	3.77	13.97	0.26
P07	32D	411	1.86	1.47	1.67	3.44	6.56	0.53
P08	32E	561	2.54	1.96	1.30	2.58	9.07	0.29
P09	36DD	570	1.64	3.35	0.51	3.29	8.82	0.37

P10	46DD	1,029	3.20	8.82	0.36	2.26	17.79	0.13
Mean	–	**603**	**1.91**	**2.64**	**1.01**	**3.14**	**10.32**	**0.37**
St. Dev.	–	**284**	**0.89**	**2.42**	**0.53**	**0.85**	**5.09**	**0.17**

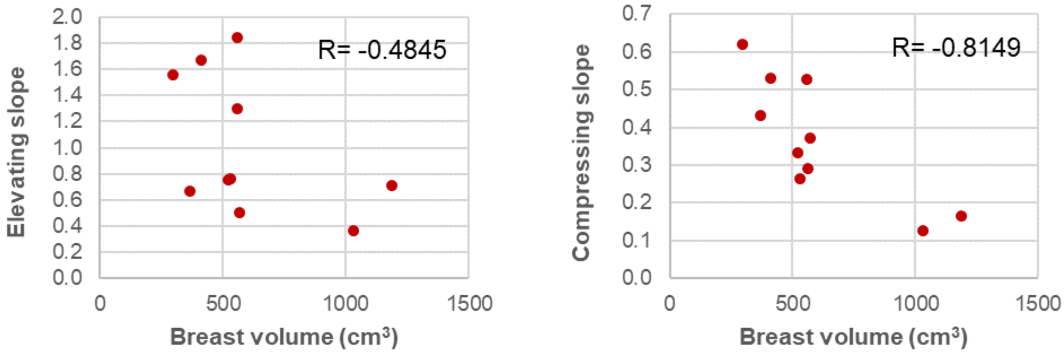

Figure 6. Correlation with breast volume: elevation slope (*tan α*, left), compression slope (*tan β*, right)

After passing the neutral point, the breast started to deform. The average force of 3.14 N was involved to press the breasts and the bust points moved for 10.32 mm by the force (**Table 2**). Similar to elevation, larger breasts were found to experience more compression, which was evidenced by a very strong positive correlation with the coefficient as high as 0.8863 between breast volume and compression distance (h_2). On contrast, compressing force (F_2) did not have any close relationship with breast volume, showing the correlation coefficient as low as -0.2802. This could also be seen as the effect of breast composition in large breasts, which have more fat percentages [2, 3, 5]. In general, adipose tissues are known to possess lower elastic modulus than glandular and fibrous tissues [27]. Therefore, it makes sense that it is easier to suppress large breasts than small breasts.

Unlike elevation, the linear relationship was well maintained between the force and displacement during compression (**Figure 3** and **Figure 5**). It took much less force, 0.37 N on average, to press the breast for a unit length (**Table 2**). This linear relationship is supported by *Hooke's Law*, where the force required to stretch/compress an elastic material is directly proportional to the extension/compression of the material. According to *Hooke's Law*, the compression slope (*tan β*) represents the spring constant of the breasts, which equals to 366.4 N/m on average. The correlation coefficients between breast volume and *tan β* was very strongly negative and -0.8149 (**Figure 6**).

The unit force for compression was associated more strongly with the breast size than the unit force for elevation; the correlation coefficients were -0.8149 for compressing while it was -0.4845 for elevating, respectively. It might be because of the severe sling slippages addressed earlier and/or the non-linear relationship between displacement and force during breast elevation (**Figure 3** and **Figure 5**). The slippages were no longer actively observed after breast sagging revoked and reliable linear slopes were able to estimate during breast compression. The empirical data indicates that compressive behaviors of breasts are very strongly correlated with the breast volume in a negative manner, which means the larger breasts are easier to deform than the smaller breasts.

3.2 *Apparent Tensile/Compression Modulus (E)*

Tensile or compression modulus defines the relationship between pulling or compressive stress and strain, as described in Equation 1. Under the current experimental configurations where breast elevation precedes prior to breast compression (**Figure 2**), the initial length (l_0) and displacement (Δl) are defined as:

$$l_0 = r + h_1 \tag{3}$$
$$\Delta l = h_2 \tag{4}$$

where r is breast radius, h_1 is elevated height, and h_2 is compressed height.

Therefore, considering the compressive height (h_2) and force (F_2) associated with the breast compression, the apparent compression modulus of breast can be further expressed as follows:

$$E = \frac{Fl_0}{A\Delta l} = \frac{F_2(r + h_1)}{A_c h_2} \tag{5}$$

where F is the force which acts, l_0 is initial length, A is the area on which the force acts, Δl is displacement, F_2 is compressing force, r is breast radius, h_1 is elevated height, A_c is cross-sectional area of a breast, and h_2 is compressed height.

Table 3 lists the resulting apparent compression modulus (E) as well as the other dimensional information associated with the calculation of apparent compression modulus for each participant.

Table 3. Apparent compression modulus of breasts and relevant measurements

	volume	θ	r	A_c	h_1	l_0	h_2 (Δl)	F_2	E
	cm^3	°	mm	cm^2	mm	mm	mm	N	N/m^2
P01	1,185	11.5	40.92	58.12	4.35	45.27	19.52	3.24	1,298.48
P02	559	8.0	46.20	45.09	1.50	47.70	6.20	3.28	5,590.31
P03	296	8.5	60.06	26.75	1.10	61.16	8.08	4.89	14,151.08
P04	520	13.5	49.95	42.62	1.58	51.53	9.08	2.93	4,018.01
P05	367	5.5	48.34	30.12	1.07	49.41	4.15	1.79	7,069.06
P06	530	11.0	34.31	35.54	1.18	35.49	13.97	3.70	2,643.74
P07	411	3.0	44.52	37.12	1.47	45.99	6.56	3.44	6,570.07
P08	561	3.5	50.10	46.38	1.96	52.06	9.07	2.58	3,274.31
P09	570	4.5	46.22	38.05	3.35	49.57	8.82	3.29	4,853.34
P10	1,029	19.0	38.16	63.09	8.82	46.98	17.79	2.26	945.30
Mean	**603**	**8.80**	**45.88**	**42.29**	**2.64**	**48.52**	**10.32**	**3.14**	**5,041.37**
St. Dev.	**284**	**5.06**	**7.16**	**11.49**	**2.42**	**6.44**	**5.09**	**0.85**	**3,807.56**

The apparent compression modulus of breast was calculated to 5.04 kN/m^2 on average and ranged from 0.95 to 14.15 kN/m^2. This is a significantly lower value compared to the tensile modulus of other soft tissues from literature (**Table 1**). For example, Krouskop *et al.* [27] tested the elastic responses of breast tissues against compression and reported the tensile modulus of 18-22 kN/m^2, 28-35 kN/m^2, and 96-116 kN/m^2 for breast fat, glandular, and fibrous tissues, respectively. Considering that an individual breast consists of varying amounts of fat, glands, and fibers, the apparent modulus looks indeed low. Future more, with the tensile modulus of breast skin reported to be much higher (~500 kN/m^2) than fat, glands and fibers [3], the apparent modulus seems unacceptably low. One of the reasons to have low apparent elastic modulus might be the over-estimated cross-sectional area (A_c) that was adopted in this research. As live human breasts are dome-shaped, and the cross-sectional area was estimated at the greatest area.

With the apparent compression modulus varying from 0.95 to 14.15 kN/m^2, the experimental data supported the previous report that the tensile modulus of breasts was extremely diverse. According to Sarvazyan *et al.* [23], the tensile modulus of breast fat was especially wider in range (5-50 kN/m^2) than the fat elsewhere (15-25 kN/m^2). The wide variability of mechanical properties could be also confirmed by the fact that the percentage of fat in the total breast volume varied in the range from 7 to 56% [2].

3.3 *Apparent Shear Modulus (G)*

Since tensile/compressive stress does not fully explain the upward motion to press breasts above the neutral position, the researchers decided to take a different approach and shear modulus (G) was selected to characterize the relationship between force and displacement associated with breast dynamics (Equation 2). The fact that the breast changes were not exactly following unidirectional extension nor contraction also supported this decision to switch the approach from tensile/compressive to shear viewpoints.

Although both values were calculated from the same empirical data, the tensile/compression and shear moduli were based on different dimensional aspects of the breasts: the former was acquired from the cross-sectional area and breast radius, while the latter was based on the breast base area and depth (**Figure 2**). Under the current experimental configurations, the initial length (l_0) and shear displacement (Δx) equals to breast depth (d) and compressive height (h_2), respectively:

$$l_0 = d \tag{6}$$

$$\Delta x = h_2 \tag{7}$$

Therefore, considering the height (h_2) and force (F_2) associated with the breast compression, the apparent shear modulus of breast can be further expressed as follows:

$$G = \frac{F l_0}{A \Delta x} = \frac{F_2 d}{A_b h_2} \tag{8}$$

where, F is the force which acts, l_0 is initial length, A is the area on which the force acts, Δx is displacement, F_2 is compressing force, d is breast depth, A_b is breast base area, and h_2 is compressed height.

Table 4 lists the resulting apparent shear modulus as well as the other measurements associated with the apparent shear modulus for each participant.

Table 4. Apparent shear modulus of breasts and relevant measurements

	volume	θ	d (l_0)	A_b	h_2 (Δx)	F_2	G
	cm^3	°	mm	cm^2	mm	N	N/m^2
P01	1,185	11.5	53.03	229.30	19.52	3.24	385.61
P02	559	8.0	47.29	183.66	6.20	3.28	1,359.81
P03	296	8.5	38.33	165.95	8.08	4.89	1,430.99
P04	520	13.5	47.51	168.84	9.08	2.93	935.33
P05	367	5.5	41.35	155.94	4.15	1.79	1,142.41
P06	530	11.0	41.80	252.08	13.97	3.70	439.04
P07	411	3.0	47.22	164.77	6.56	3.44	1,520.83
P08	561	3.5	47.53	158.22	9.07	2.58	876.26
P09	570	4.5	44.35	130.02	8.82	3.29	1,269.07
P10	1,029	19.0	54.40	250.17	17.79	2.26	276.04
Mean	**603**	**8.80**	**46.28**	**185.89**	**10.32**	**3.14**	**963.54**
St. Dev.	**284**	**5.06**	**5.03**	**42.59**	**5.09**	**0.85**	**459.67**

Having 0.96 kN/m^2 on average, the apparent shear modulus was still estimated lower than the reference values from literature (**Table 1**). Geerligs, Peters, Ackermans, Oomens, and Baaijens [34] reported 7.5 kN/m^2 for adipose tissues, while Sun *et al.* [32] suggested the range of 4.0 kN/m^2. A similar reason is counted for the low moduli that the cross-sectional areas (A_b) were over-estimated for the dome-shaped human breasts by measuring it at the greatest area.

Unfortunately, unlike tensile/compression modulus, there was no literature investigating shear modulus of breast tissues. The apparent shear modulus observed in this research ranged from 0.28 to 1.52 kN/m^2 that was as wide as the tensile/compression modulus. It was found that there was a strong positive correlation (R=0.77) between apparent tensile/compression and shear moduli.

3.4 *Apparent Elastic Properties of Human Breast*

Both apparent tensile/compression and shear modulus showed a strong negative relationship with breast volume (**Figure 7**), but the shear modulus had slightly stronger correlation than the tensile modulus. According to Krouskop *et al.* [27], breast fat showed significantly lower tensile modulus (18-22 kN/m^2) than glandular (28-35 kN/m^2) and fibrous (96-116 kN/m^2) tissues. Since larger breasts are known to include more adipose tissues than smaller breasts [2, 3, 5], high breast volume leads to low apparent moduli, which indicates those breasts are more elastic. This explains why large-breasted females suffer from exercise-induced breast discomfort [18]. Due to their low apparent modulus, large breasts would undergo more deformation when the same amount of force is induced.

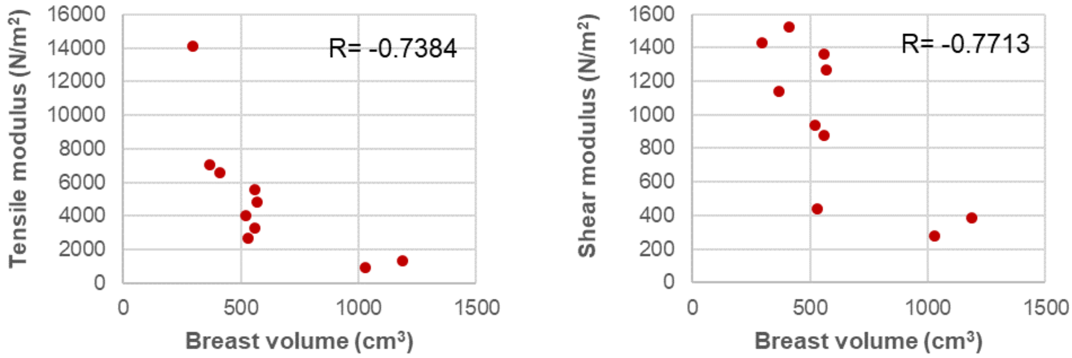

Figure 7. Correlation with breast volume: apparent tensile modulus (left), apparent shear modulus (right)

Thinking in depth about which approach could characterize the apparent elastic behaviors of human breasts more rationally, the impact of breast shape was brought into consideration. The equation for tension/compression modulus (Equation 5) suggests that, if two breasts of the same volume have the exact same apparent compression modulus, it would take less force (F_2) to compress the one in a flat shape, which has relatively smaller cross-sectional area (A_c) and larger breast radius (r) than the one in a protruding shape. On the other hand, the equation for shear modulus (Equation 8) advises the opposite in which the protruding breast with smaller base area (A_b) and larger breast depth (d) could show more apparent elasticity, requiring less force to press.

It would make more sense to explain breast apparent elasticity from the shear modulus perspective. This aligns with how compression sport bras work, where breast bounces are controlled by flattening the breasts [9, 18, 35]. By increasing base area (A_b) and reducing breast depth (d), one could reduce the breast apparent elasticity and therefore less breast deformation would take place. This leads to the emerging necessity to investigate soft tissues and materials for their shear modulus.

4. Conclusion

Applying two different theoretical models of elasticity, tensile/compression and shear moduli, the current research explored apparent elastic properties of human breasts. The apparent elastic behavior of breasts was observed through force-displacement measurements in a laboratory setting with the consideration of individual breast geometry.

Since destructive tests and sharp clamping were never feasible with live breast measurements, it was impossible to follow the standard procedures for tensile and shear tests. In consequence, there are several limitations associated with research methods in terms of the lack in accuracy, sensitivity, and generalizability. The research findings cannot be applied to explain the apparent elasticity of breasts directly because of the limited sample size. Another limitation to be noted is relatively young populations involved in the experiments. Since the subjects were recruited by convenient sampling under the university environments, the average age of the subjects was as low as 20.4. Therefore, the research findings could be different for other generations. Specifically, the lower apparent elastic modulus is expected with more aged populations.

Despite the restrictions in experimental measurements, the research conclusion offers meaningful insights on breast apparent elasticity. The research findings suggested choosing the shear approach over tensile or compressive approach to characterize the apparent elasticity of live human breasts. With further investigations, the apparent shear modulus of breasts can successfully conceptualize how to understand the mechanics of human breasts. Future research must include breast displacements in medial/lateral and posterior/anterior directions. By characterizing breast apparent elasticity, it would be possible to define how encapsulation (*e.g.* bra cups) and harnesses (*e.g.* shoulder straps) increase breast stiffness and engineer the mechanical properties of bra components.

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Conflict of interest

There is no conflict of interest for this study.

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