

# Mechanics of the normal woman's breast

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**Abstract.** Knowledge on the forces acting on a woman's breast and on the mechanical properties of the breast tissues is important for studying the effects of plastic surgery techniques for breast reconstruction as well as for the design of cosmetic breast implants. Surprisingly, there are no data in the literature regarding mechanical loads on the breast tissues during daily or sport activities, and there are no coherent sources of data in regard to mechanical properties of the breast tissues. Accordingly, this paper is aimed at reviewing the mechanics of the normal breast. First, the anatomy of the breast and major aging-related changes are described. Second, the mechanical characteristics of all tissue components of the breast are detailed. Last, analytical approximations are made in regard to the forces acting on the breast during normal activity, and the respective internal breast forces supported by the suspensory ligaments, pectoralis fascia and ribs are calculated. The data presented in this paper are useful for biomechanical modeling of the breast as well as for evaluating the loads acting on surgical repairs and breast implants.

Keywords: Plastic surgery, implant design, biomechanical model, tissue mechanical properties

## 1. Introduction

Knowledge on the forces acting on a woman's breast and on the mechanical properties of the breast tissues is important for studying the effects of plastic surgery techniques for breast reconstruction as well as for the design of cosmetic breast implants. Surprisingly, there are no data in the literature regarding mechanical loads on the breast tissues during daily or sport activities, and there are no coherent sources of data in regard to mechanical properties of the breast tissues. Accordingly, this paper is aimed at reviewing the mechanics of the normal breast. First, the anatomy of the breast and major aging-related changes are described. Second, the mechanical characteristics of all tissue components of the breast are detailed. Last, analytical approximations are made in regard to the forces acting on the breast during normal activity, and the respective internal breast forces supported by the suspensory ligaments, pectoralis fascia and ribs are calculated. The data presented in this paper are useful for biomechanical modeling of the breast as well as for evaluating the loads acting on surgical repairs and breast implants.

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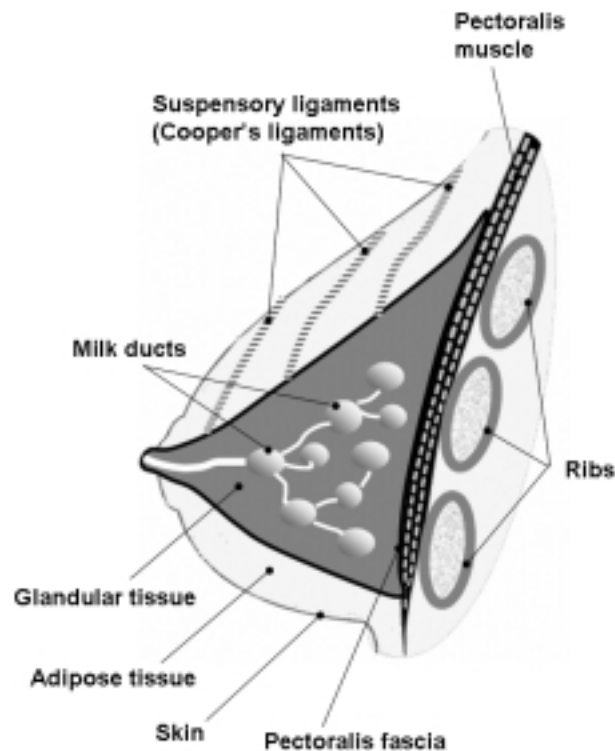


Fig. 1. Anatomical scheme of the mature pre-menopausal breast, showing the structure and location of the suspensory ligaments (of Cooper).

## 2. Anatomy of the breast

A woman's breasts sit over the pectoralis major muscle and usually extend from the level of the second rib to the level of the sixth rib anteriorly. The breasts cover a large part of the chest wall. In front, the breast tissue may extend from the clavicle (collarbone) to the middle of the sternum (breastbone). On the side, breast tissue may continue into the axilla (armpit) and reach as far as the latissimus dorsi (muscle extending from the lower back to the humerus bone of the upper arm). The breast is an inhomogeneous structure containing different tissue layers (Fig. 1), however, the two predominant types of tissue within the breast are fat and glandular tissue, which supports lactation.

The dimensions and weight of the breast can vary substantially between individuals. A small to moderate breast weighs about 500 g or less [36], and large breasts weigh about 750 to 1000 g [19]. Some women have more glandular tissue in their breasts and some have less, and likewise, some have more fatty tissue or connective tissue than others, and the ratio of fat to connective tissue content determines the firmness of the breast. The size and shape also varies over time in the same woman because of the changes during menstrual cycle, pregnancy, after weaning, and during menopause [3].

The mammary gland forms a cone with its base at the chest wall and its apex at the nipple (Fig. 1). The superficial layer (fascia) is separated from the skin by 0.5–2.5 cm of subcutaneous fat. Tentacle-like prolongations of fibrous tissue extend in all directions from this fascia to the skin; these are called the suspensory ligaments, or Cooper's ligaments [3]. In the adult mammary gland, there are 15–20 irregular lobes, converging to the nipple through ducts 2 to 4.5 mm in diameter [3]. These ducts are immediately surrounded by dense connective tissue, which acts as a supporting framework [3]. The glandular tissue

Table 1  
Mechanical properties of tissue components of the breast

Tissue type	Elastic modulus [kPa]	Ultimate strength [MPa]
Ribs	2,000,000–14,000,000	100
Pectoralis major and minor muscles	<i>In the longitudinal direction:</i> For dynamic loading: ~30 <i>In the transverse direction:</i> For dynamic loading: 1.5–6 For static loading: 0.75–3.6	0.4–0.7
Pectoralis fascia	100–2000	20–100
Suspensory ligaments	80,000–400,000	40
Glandular tissue	7.5–66	No data available
Adipose	0.5–25	No data available
Skin	200–3000	20

\*References to mechanical property data are provided in the text.

\*\*Mechanical properties provided herein should be considered as representing young healthy individuals; it should be considered that these tissue properties may be affected by age and disease.

is supported by estrogen. When a woman reaches menopause the estrogen levels decrease and the glandular tissue atrophies and eventually disappears, leaving only fat, superficial fascia, the suspensory ligaments and skin [3]. The structural supporting system of the breast, made of the superficial fascia, the suspensory ligaments and skin, can also change its fibrous frame with aging and due to the force of gravity. Specifically, it had been reported that breast ptosis (“dropping”) is always related to the laxity of the superficial fascia, suspensory ligaments and skin. A comprehensive review of the anatomy of the fibrous frame of the breast was published by Riggio and colleagues [38].

### 3. Mechanical properties of the breast tissues

Mechanical properties of tissues contained in the breast, or which support the breast structure (Fig. 1) are essential for any biomechanical modeling attempt, as well as for the design of biomechanically-compatible breast implants. In particular, the elastic moduli and strength properties of the tissues involved are of interest. Mechanical properties of breast tissues were mainly studied for adipose, glandular tissue and skin. All soft tissues of the breast can be assumed to be nearly incompressible, i.e. with a Poisson's ratio of  $\sim 0.5$ . For tissues other than adipose, glandular tissue and breast skin, approximations of the mechanical properties can be made based on mechanical properties of human tissues with similar composition or structure, for which mechanical properties were reported in the literature. Accordingly, an up-to-date review of the mechanical properties of breast tissues is provided below, and in Table 1.

#### 3.1. Ribs

The second to sixth ribs serve as the rigid structural support to the breast (Fig. 1). Accordingly, the rib response to dynamic loading is of particular interest in the design of breast implants. Previous work on human ribs (2<sup>nd</sup>, 4<sup>th</sup>, 6<sup>th</sup> and 8<sup>th</sup>) included non-destructive mechanical tests performed using a cantilever end condition, which resulted in force-displacement data showing no difference in stiffness among the ribs tested at different levels of the thorax [43]. However, cantilever end condition is not an accurate way to reproduce physiological rib loading. Another study performed static 3-point bending tests on lateral rib sections from the 6<sup>th</sup> and 7<sup>th</sup> ribs [18]. For the ten cadavers tested, loading in the

lateral/medial direction resulted in an average failure stress of 106 MPa and an average Young's modulus of 11.5 GPa [18]. The same authors later published data from static 3-point bending tests on a total of 79 male cadaver subjects, utilizing the 6<sup>th</sup> and 7<sup>th</sup> ribs from each [46]. Quasi-static loading was performed in the medial/lateral direction at rates of 2.54, 0.508 and 12.7 mm/min; the average failure stress for all tests was 100.68 MPa, with a variance of 19% of the mean [46]. More recently, quasi-static 3-point bending tests were performed on the 7<sup>th</sup> and 8<sup>th</sup> ribs of 30 subjects [56], and these authors, similarly, found no significant difference between the strength parameter for the two ribs. They reported, however, that there was a mild difference in the average force at fracture of the 7<sup>th</sup> and 8<sup>th</sup> ribs, that is, 153 and 137 N, respectively [56]. An average elastic modulus of 2.3 GPa for the 7<sup>th</sup> rib and 1.9 GPa for the 8<sup>th</sup> rib was determined, however, these values were obtained from a relatively slow loading rate, of 2.5 mm/min [56]. The difference between elastic moduli reported in [18] and [56] can be attributed to the viscoelastic response of the bone.

Most recently, tensile elastic moduli and strength values were measured in 117 cortical bone specimens from ribs 1–12 of 6 cadavers (3 males and 3 females, age 18–67 years) [24]. The overall average of all cadaver data gave an elastic modulus of 13.9 GPa, a yield stress of 93.9 MPa, a yield strain of 0.883%, an ultimate stress of 124.2 MPa, an ultimate strain of 2.7%, and a strain energy density of 250.1 MPa-strain [24]. The author of [24] noted that the rib cortical bone became more brittle with increasing age, shown by significant increase in the modulus and decrease in peak strain. Taken together, the published results indicate that there is no significant difference between mechanical properties of ribs at different levels of the thorax, that rib elastic moduli (rib cortex alone, and whole rib data pooled) are in the range of 2–14 GPa, and that ultimate strength is about 100 MPa.

### 3.2. *Pectoralis major and minor muscles*

The breasts overlay vital chest wall muscles such as the pectoralis major (the 'pecs'), the pectoralis minor (thin, triangular muscle beneath the pecs), and the intercostals (muscles between the ribs) (Fig. 1). The breasts also may cover some of the serratus magnus (also called the serratus anterior; a slender muscle that is attached to the ribs/rib muscles and connects with the shoulder blade) and the rectus abdominis (long, flat muscle that stretches up the torso from the pubic bone to the ribs). Depending on the body posture, the weight of the breast may induce static or dynamic shear forces (when standing or walking), compression (when lying supine), or tension (when kneeling on four) on the pectoralis major and minor muscles.

Very little information is available in the literature regarding passive viscoelastic mechanical behavior of living or fresh muscle tissue *from humans*. Specifically, there is paucity of data regarding directional *transverse* elastic or shear moduli for living/fresh human muscles (since in some postures, e.g. lying, forces act perpendicularly to the direction of muscle fibers). *In vitro* tensile tests of fresh human muscle fibers incidentally extracted during surgical treatment of trauma (e.g. from the brachioradialis, thumb and wrist extensors) showed that the longitudinal elastic modulus was  $28 \pm 3$  kPa (mean  $\pm$  standard deviation) [14]. Considering that muscle tissue contains mostly water ( $\sim 75\%$ ) and therefore, is nearly incompressible (Poisson's ratio approaches 0.5), shear moduli of human muscle tissue can be approximated from  $G = E/2(1 + \nu)$ , as  $\sim 9$  kPa. *In vivo* measurements obtained with ultrasound-based or MR-based elastography are in good agreement with this estimate. Shear moduli of unloaded human muscle tissue obtained using elastography for the lateral gastrocnemius [6,48], biceps brachii [12], flexor digitorum profundus [48] and soleus [48] are in the ranges of 5 to 20 kPa, and standard deviations are around 5 to 8 kPa. Only one paper used elastography to study *in vivo* directional properties of muscle

tissue [17], and found that the transverse shear modulus is in the order of 1/10 the longitudinal modulus. Accordingly, the above studies (taken together) suggest that the transverse instantaneous shear moduli of human skeletal muscles (that is, that are relevant for dynamic body activities) are in the order of 500 Pa to 2 kPa. No *in vivo* stress relaxation tests of skeletal muscles under transverse loading are available from humans, but stress relaxation in transversally loaded living tibialis anterior of rats showed that long-term relaxation forces are in the order of 0.5 to 0.6 the instantaneous force [9], and this was recently confirmed in stress relaxation tests obtained by Gefen and colleagues [15]. Because shear moduli are linearly proportional to the relaxation force [15], it can be deduced that *long-term transverse* shear moduli of human skeletal muscles (which are relevant for static body postures) should be in the order of 250 Pa to 1200 Pa. Assuming incompressibility of muscle tissue, long-term elastic moduli in the transverse direction should be in the order of 750 Pa to 3600 Pa. Corresponding instantaneous elastic moduli (which are relevant to dynamic body activities) should be in the order of 1500 to 6000 Pa. Tensile strength of fresh muscle tissue was not reported for human specimens, but in rat soleus and rectus femoris, it was found to be at the range of 0.4–0.7 MPa [25].

### 3.3. *Pectoralis fascia*

The pectoralis fascia is a thin lamina, covering the surface of the pectoralis major muscle, and sending numerous prolongations between its fasciculi. It is attached, in the middle line, to the front of the sternum. Above, the pectoralis fascia attaches to the clavicle. Laterally and below, it is continuous with the fascia of the shoulder, axilla, and thorax. No published data exist on the mechanical properties of the pectoralis fascia, however, mechanical properties of muscle fascia from the legs had been studied. Specifically, *in vivo* measurements of elastic modulus of muscle fascia were reported to be ~530–1200 MPa for the tibialis anterior muscle [21,28], and ~1100–1800 MPa for the triceps surae muscles [30]. Fascia tissue in the foot, however, was assigned lower elastic moduli, of ~90–350 MPa [16]. Strength of fascia tissue was studied for plantar fascia as well, and was found to be in the range of ~20–30 MPa [55]. Strength was evaluated, but not directly measured for the fascia of the triceps surae, and was reported to be about 100 MPa [30]. Considered together, the above literature allows only first approximation of the mechanical properties of the pectoralis fascia, since this specific tissue has not been studied directly. Nevertheless, it is reasonable to assume that the elastic modulus of pectoralis fascia is in the range of 100–2000 MPa, and the ultimate strength in the range of 20–100 MPa.

### 3.4. *Suspensory ligaments*

The subcutaneous fat layer in the breast is crossed by thin ligaments called the suspensory ligaments or Cooper's ligaments (Fig. 1). They run oblique to the skin surface, from the skin to the deep pectoral fascia [3]. Ligaments are composed of closely packed collagen fiber bundles oriented in a parallel fashion to provide for structural stability [3]. The major cell type is the fibroblast and they are interspersed in the parallel bundles of collagen. While there are no specific experimental data for mechanical properties of suspensory ligaments of the breast in the literature, their properties can be extrapolated from those known for other ligamentous structure in the human body. For example, in knee ligaments, between 65 and 70% of a ligament's total weight is composed of water. On a fat-free basis, Type I collagen is the major constituent (70–80% dry weight) and is primarily responsible for a ligament's tensile strength. Type III collagen (8% dry weight) and Type V collagen (12% dry weight) are other major components [54]. Tensile elastic moduli of knee ligaments are reviewed in detail in [54], and are reported to be in the range of 80–400 MPa. Tensile strength was reported to be at about 40 MPa [35,54]. Older age (50 years

old and over) was shown to decrease the elastic modulus of knee ligaments by  $\sim 40\%$ , and the tensile strength by  $\sim 65\%$  [35]. Overall, though mechanical properties of the breast's suspensory ligaments were not studied directly, first approximation of their elastic moduli and strength can be made based on other ligament properties, as 80–400 MPa and 40 MPa, respectively.

### 3.5. Glandular tissue

The glandular tissue of the breast (Fig. 1) houses the lobules (milk producing glands at the ends of the lobes) and the ducts (milk passages). Toward the nipple, each duct widens to form a sac (ampulla). During lactation, the bulbs on the ends of the lobules produce milk. Once milk is produced, it is transferred through the ducts to the nipple. Quantitative elastograms showed heterogeneity in the shear moduli of breast tissues, generally reporting stiffer values in glandular tissues than in adipose tissue. In the Saravazyan et al. paper [42], normal glandular tissue is reported to be 5-fold to 50-fold stiffer than adipose. In the Wellman's [52] and Samani et al.'s [41] studies, glandular tissue was found to be 6.7 times stiffer than adipose. In the Krouskop et al. data 4.5 [26]; in McKnight et al. data 2.3 [33]; in Bakic model 1.2 [5], and in the Azar model 1.0 [4]. Values of elastic moduli reported for glandular tissue range between 2 and 66 kPa, depending on the magnitude of strains (up to 20%) and on the testing method [26, 33,45]. Consistently, finite element modeling studies of breast tissue used values in the order of 10 kPa for the elastic modulus of glandular or fibroglandular tissue [39,40]. Strength properties for glandular tissue are not yet available in the literature.

### 3.6. Adipose

The white form of adipose tissue, which is the type present in the breast (Fig. 1), mostly contains lipidic fluid (60–85% weight), which compose of 90–99% triglycerides, as well as free fatty acids, diglycerids, cholesterol phospholipids and minute quantities of cholesterol ester and monoglycerides. The remaining components are water (5–30% weight) and protein (2–3% weight) [1]. Much of the published information on the mechanical properties of white adipose tissue concerns specimens from the human breast [41,50]. Using mechanical testing and elastography, the instantaneous shear modulus for adipose tissue from breast samples was found to be in the range of 0.5–25 kPa [27,33,41,45,50]. Adipose is more liquefied at 37°C than at room temperature, and so, elastic modulus of breast adipose tissue is probably lower *in vivo* than when tested *in vitro* at room temperature. For tissue strains higher than  $\sim 15\%$ , the modulus of adipose tissue stiffens and becomes more similar to that of glandular tissue [3]. At this time, there is paucity of information in the literature regarding the strength properties of adipose tissue.

### 3.7. Skin

The skin consists of three layers: the epidermis, dermis, and hypodermis. The epidermis thickness ranges between 50 to 100  $\mu\text{m}$ . It is composed of a "dead" layer of cells called "stratum corneum" (10 to 20  $\mu\text{m}$  thick on most of the body surface). These cells are flat and mainly composed of keratin, a quite rigid and hard material. The stratum corneum forms a protective shield for covering the underlying viable epidermis composed primarily of keratinizing epithelial cells. The dermis, depending on the anatomical sites, is mainly composed of collagen and elastin fibers embedded in a viscous medium made of water and glycoproteins. Fibers of the upper dermis (or "papillary dermis") are thinner than those present in the deep dermis. Total thickness varies according to the site from 1 to 3 mm. Hypodermis is

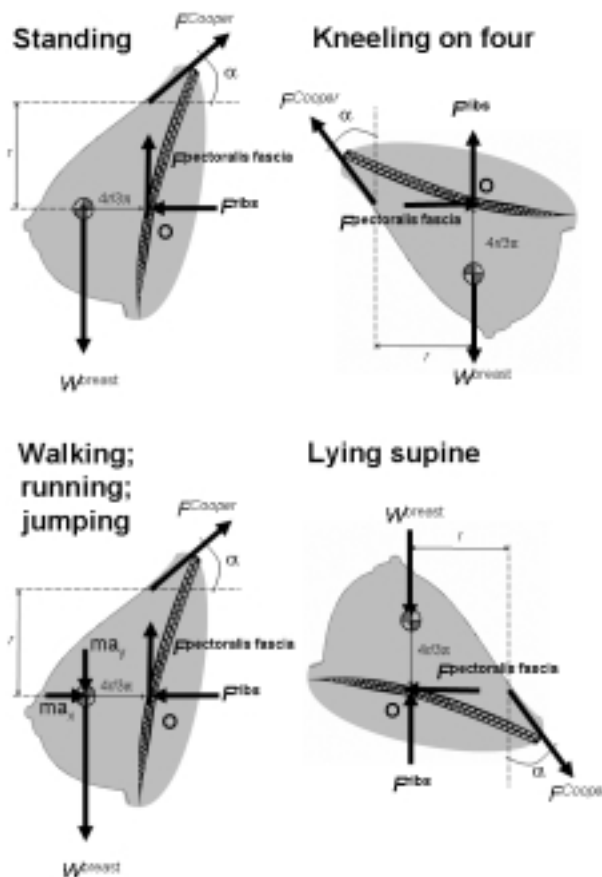


Fig. 2. Diagrams showing the internal forces acting on the breast tissues during static postures and dynamic activities.  $r$  = effective radius of curvature of the breast;  $\alpha$  = dorsal insertion angle of the breast; O = reference point at the base of breast.

quite variable in thickness depending on the person, and the position on the body; it is mainly composed of cells (adipocytes).

Human skin mechanical properties are highly non-linear, viscoelastic and anisotropic, and vary with age, hydration, obesity, disease and anatomical site. However, skin of the breast can be considered as a linear isotropic material at strains less than 50% [40]. The aging process considerably alters both the structure and the mechanical properties of skin. Aged skin is less extensible and less elastic than adult skin [13]. These alterations could be related to important modifications occurring in the upper dermis where fibers are markedly modified, thinned and/or fractionated as revealed by ultrasound imaging [37] and ultrastructural [7] studies. Changes in the mechanical properties of skin versus age have been extensively studied with noninvasive physical methods in the last decade. After a long period of debate on some conflicting results, most of the investigators now agree on the following points: (1) the elastic part in the total strain of skin (after the application of a given stress) is reduced in aged skin, (2) the total strain is reduced. This means that the elastic modulus is increased. This elastic modulus ranges from 0.2 to 3 MPa [44], and during aging it increases by about 30% [29]. In regard to strength, Holzmann and colleagues have reported *in vitro* measurements on chest skin thickness (1.9 mm above the sternum), tensile strength (19.4 MPa) and failure strain ( $\sim 60\%$ ) [20]. According to Sugihara, the extensibility slightly decreases with age on chest and anterior thigh whereas abdomen skin extensibility does not

Table 2

Internal forces in the breast tissues (corresponding to the free body diagrams of the breast in Fig. 2). Forces include tension in the suspensory (Cooper's) ligaments ( $F^{Cooper}$ ), shear in the pectoralis fascia ( $F^{pectoralis\ fascia}$ ), and compression of the ribs ( $F^{ribs}$ ). The angle  $\alpha$  denotes *the dorsal insertion angle of the breast*

Activity	Internal Breast Forces		
	$F^{Cooper}$	$F^{pectoralis\ fascia}$	$F^{ribs}$
Standing	$\frac{4W^{breast}}{3\pi \cos \alpha}$	$W^{breast} \left(1 - \frac{4 \tan \alpha}{3\pi}\right)$	$\frac{4W^{breast}}{3\pi}$
Lying supine	0	0	$W^{breast}$
Kneeling on four	0	0	$W^{breast}$
Walking; running	$\frac{4}{3\pi \cos \alpha} (W^{breast} + m^{breast} a_y)$	$(W^{breast} + m^{breast} a_y) \left(1 - \frac{4 \tan \alpha}{3\pi}\right)$	$\frac{4}{3\pi} (W^{breast} + m^{breast} a_y) + m^{breast} a_x$
Vertical jumping*	$\frac{4}{3\pi \cos \alpha} (W^{breast} + m^{breast} a_y)$	$(W^{breast} + m^{breast} a_y) \left(1 - \frac{4 \tan \alpha}{3\pi}\right)$	$\frac{4}{3\pi} (W^{breast} + m^{breast} a_y)$

\*In the vertical jumping condition, the free body diagram of walking/running is used for calculation of the internal breast forces, however, it is assumed that the horizontal component of breast acceleration ( $a_x$ ) equals zero.

seem to change significantly [47]. Vogel conducted an extensive *in vitro* research involving 348 autopsy specimens [51]. Samples were taken from the skin above the sternum; dumbbell-shaped specimens were cut along the same axis. The specimens were 50 mm long and the narrow section was 4 mm wide [51]. At the age of 30, the skin thickness above the sternum was approximately 1.75 mm, tensile strength was 20 MPa and elongation at failure was 72% [51]. Taken together, these studies indicate that human skin at the chest region is ~2 mm thick, have an elastic modulus of 0.2–3 MPa, tensile strength of ~20 MPa, and failure strain of about ~60–75%.

#### 4. Internal forces in the breast tissues

Internal forces in the breast tissues were approximated analytically using free body diagrams of the breast at static postures, namely, standing, kneeling on four, and lying supine, as well as during dynamic activities including running, stair climbing and jumping (Fig. 2). In order to allow analytical solutions, we assumed that three tissue structures of the breast support the static and dynamic body loads: Cooper's ligaments, the fascia of the pectoralis muscle, and the ribs. Cooper's ligaments support tension loads and transfer it to the dorsal skin of the breast. The pectoralis fascia supports shear at the base of the breast, and compression is supported by the rib cage. We further assumed that at the base of the breast, where the pectoralis fascia and rib reaction forces act (point "O", Fig. 2) there are no reaction moments, and that the angular acceleration of the breast during dynamic activities is small, and can be ignored. By employing these assumptions, it was possible to use equilibrium of force vectors and moments in order to derive analytical solutions for the forces transferred through Cooper's ligaments ( $F^{Cooper}$ ), the pectoralis fascia ( $F^{pectoralis\ fascia}$ ), and the ribs ( $F^{ribs}$ ) during static and dynamic activities. These internal breast forces depended on the breast weight ( $W^{breast}$ ) and mass ( $m^{breast}$ ), the dorsal angle of breast insertion  $\alpha$ , and, for all dynamic activities, on the vector of linear acceleration of the center of mass of the breast at the sagittal plane [ $a_x, a_y$ ] (Fig. 2). The terms obtained for  $F^{Cooper}$ ,  $F^{pectoralis\ fascia}$ , and  $F^{ribs}$  for each of the studied postures/activities, by means of equations of equilibrium of forces and moments in the sagittal plane, are provided in Table 2. In order to obtain approximated values of internal forces in the breast tissues (corresponding to Table 2), breast masses  $m^{breast}$  of 500–1000 g were assumed. The dorsal insertion angle of the breast  $\alpha$  was taken as 45–60°, following the anatomical charts of Riggio et al. [38].



Table 3

Trunk acceleration components during different dynamic activities, for utilization in calculation of internal forces in the breast. Maximal reported accelerations were listed (see text for references), in order to allow calculation of maximal internal breast tissue forces

Activity	Trunk acceleration components [g]	
	Vertical ( $a_y$ )	Anterior-posterior ( $a_x$ )
Walking*	0.8	0.4
Running (or jogging); stair climbing	5	2.5**
Vertical jumping – free	2	0
Vertical jumping – trampoline	6	0

\*The Mason et al. [32] data of “walking” can actually be classified as belonging in the category of sport-oriented studies (and thus, in the “running/jogging” item above) owing to the high walking speed (7 km/h) set in that study (see text).

\*\*The ratio  $a_y/a_x$  of the walking condition is assumed, in lack of published data that is specific for the trunk.

#### 4.1. Breast acceleration during dynamic activities

It is convenient to quantify accelerations of the moving human body (in  $\text{m/s}^2$ ) by normalizing data with respect to the acceleration of gravity ( $g = 9.81 \text{ m/s}^2$ ). Generally, amplitudes of the components of the body's acceleration vector during human locomotion are higher in the vertical direction than in both horizontal directions (i.e. anterior-posterior and medial-lateral) [10]. Measured acceleration values are also usually higher when acquired at inferior body parts, compared with superior body parts [10]. During walking, for instance, the accelerations of the upper body determined from stereophotogrammetry range from 0.3 to 0.8 g in the vertical direction, whereas in the horizontal directions they range from  $-0.2$  to 0.2 g at the head and from  $-0.3$  to 0.4 g at the low back [11]. During slow to fast gait, the vector of trunk acceleration peaks at push-off, with vertical and horizontal components of approximately 0.4–0.5 g and 0.2–0.3 g, respectively [22,23]. During running, Bhattacharya and colleagues observed absolute vertical peak accelerations ranging from 0.8–4.0 g at the head and 0.9–5.0 g at the low back using their skin-mounted accelerometers [8]. Stair climbing produced peak accelerations that are similar to those occurring during running [53]. During free vertical jumps of adults, fly-time accelerations of the trunk peak at  $\sim 2$  g [2]. During trampoline jumping, absolute vertical peak accelerations are higher, between 3.0–5.6 g at the head, and between 3.9–6.0 g at the low back [8,10]. Only one study, by Mason and colleagues, measured acceleration of the actual female breast, in 3 young subjects, during running, jogging, aerobics march and walking [32]. Accelerations were derived from high speed cine films (100 frames/s) taken at 5–7 s timeframes, where the nipple was used as a marker for the optical motion tracking system [32]. Running/jogging/aerobics with a bare breast caused vertical accelerations of  $\sim 2.5$ –3 g; running with normal bra reduced vertical accelerations by  $\sim 0.7$  g, and running with a sport bra reduced acceleration by  $\sim 1.2$  g [32]. Walking with a bare breast caused vertical accelerations of  $\sim 1.7$  g, whereas normal and sport bras reduced the acceleration by 0.2 g and 0.5 g, respectively [32]. It should be noted that the walking speed at the Mason et al. study was high, 7 km/h, which was set to simulate a power walking experience [32].

Based on the literature reviewed above, characteristic values of maximal vertical and horizontal (anterior-posterior) components of linear acceleration of the breast were determined, for further calculation of dynamic internal breast forces. These linear acceleration components are summarized in Table 3.

Table 4

Approximated values of internal forces in the breast tissues (corresponding to Table 2). A breast weight of 500 – 1000g was assumed. The dorsal insertion angle of the breast ( $\alpha$ ) was taken as 45–60°. Acceleration data were adopted from Table 3

Activity/Internal Breast Forces	$F^{\text{Cooper}}$ [N]	$F^{\text{pectoralis fascia}}$ [N]	$F^{\text{ribs}}$ [N]
Standing	2.9–8.3	1.3–5.6	2.1–4.2
Supine	~ 0	~ 0	4.9–9.8
Kneeling on four	~ 0	~ 0	4.9–9.8
Walking	5.3–15	2.3–10.2	5.7–11.4
Running; stair climbing	17.7–50	7.8–33.9	24.8–49.5
Vertical jumping – free	8.8–25	3.9–16.9	6.2–12.5
Vertical jumping - trampoline	20.6–58.3	9.1–39.5	14.6–29.1

#### 4.2. Calculation of static and peak dynamic tissue load values

By substituting a normal range of breast weights (500–1000 g [36,19]), a range of dorsal insertion angles of the breast  $\alpha$ (45–60° [38]), and the acceleration magnitudes of Table 3 in the force equations provided in Table 2, analytical approximations of static and peak dynamic tissue load values can be obtained. This analysis is provided in Table 4. The results in Table 4 indicate that during standing, walking, running and jumping, the maximal soft tissue forces are carried by the suspensory (Cooper's) ligaments, which is reasonable, considering that these ligaments are considered (anatomically) as the structural framework which holds the breast [38]. The breast deforms with age so that its dorsal insertion angle  $\alpha$  decreases [38]. This implies that the forces transferred through the suspensory ligaments to the dorsal skin also decrease with age (i.e. proportionally to  $1/\cos\alpha$  Table 2). Importantly, Table 4 indicates that (1) maximal forces in Cooper's ligaments occur during running (or stair climbing), and jumping, and that (2) during strenuous activity, such as running or jumping on a trampoline, Cooper's ligaments may be subjected to transient loads with peaks that are as much as 50–60 N.

#### 4.3. Cyclic loading characteristics for the breast tissues

Loading of the breast tissues in “real world” conditions is a combination of static and dynamic loading periods. For example, a study conducted on Dutch operation room nurses revealed that their average standing time per day is 2.5 hours, however, some subjects reported that they stand 2/3 of their workday time [34]. Yet, since maximal breast loads occur during dynamic activities (Table 4), with walking being the most common activity, the number of steps per timeframe appears to be a better measure for evaluating the loading scenario for normal breasts. There is limited information about the number of steps taken on a daily or yearly basis. However, a small sample study (in 6 healthy individuals) revealed that about 5000 steps are made per day [31]. Extrapolation of that study indicates that the number of steps per year would be on the order of 2 million [49]. Hence, it is reasonable to assume that the healthy breast is subjected to cyclic forces with peaks of 5–15 N (depending on the individual breast size), 5000 times per day or ~2 million times per year of normal activity (Table 4).

### 5. Summary

Plastic surgery of the breast has evolved substantially over the last years. Surgical interventions to reconstruct or reshape the breast are needed after mastectomy (removal of masses in the breast that are suspected as cancerous) and for cosmetic purposes, e.g. augmentation mammoplasty (change of the

breast size) or mastopexy (breast “lifting”). Breasts are rebuilt with implants or with autologous tissue flaps (excess fat and sometimes muscle), and in some cases, implant and flap techniques are combined. Regardless of the technique being used to reconstruct or reshape the breast, the implant or flap must be able to support the mechanical forces acting on the breast when the patient returns to normal activity. Otherwise, mechanical failure of the implant or necrosis of the biological flap may occur. Surprisingly, there are no published data in the literature regarding the magnitudes of forces acting on the breast, even for healthy breasts subjected to normal activity profiles.

Information on the mechanical forces acting on the normal breast is essential for the design and evaluation of new breast implants or surgical flap techniques. This is the first paper to provide such data, and though only first approximations, based on relatively simple engineering calculations are made here, data can serve as a baseline for future research. Specifically, this paper indicates that the most highly loaded soft tissue structure in the breast, either during static body postures or during dynamic activities, is the suspensory ligament system (Fig. 1). During static standing, forces on the suspensory ligaments may be up to 10 N, whereas during walking, running and jumping, these forces rise to be on the order of 15, 50 and 60 N, respectively (Table 4). The review of mechanical properties of breast tissues and the loading conditions on these tissues during different postures/activities that are provided herein are also a solid base for future finite element analysis studies, to determine local tissue loading, namely, stress and strain distributions, in the breast. This will allow bioengineers to determine the biomechanical interaction of breast implants with the breast tissues, in an objective and quantitative manner.

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