Development of a handsfree speech valve for laryngectomy patients

Ward van der Houwen

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Promotores:	Prof. dr. ir. G.J. Verkerke Prof. dr. B.F.A.M. van der Laan
Beoordelingscommissie:	Prof. dr. G. Rakhorst Prof. dr. ir. A.O. Eger Prof. dr. F.J.M. Hilgers

Voor mijn lieve en wijze oma Miep

Contents

Preface

The department of Biomedical Engineering, section Artificial Organs Research, headed by Prof. G. Rakhorst, has a long history in the field of researching voice restoration. The department has housed highly esteemed phoniatrists like Prof. H.K. Schutte and Dr J. Sveč. A long-standing relation exists between our department, several Ear Nose and Throat (ENT) departments throughout Europe, and industries in speech rehabilitation. This thesis is an example of some of the work done in this multidisciplinary field. The majority of studies published here were conceived in close co-operation with Ph.D. student T.J. van Kalkeren, MD, and resident at the UMCG department of Otorhinolarygology/Head and Neck Surgery, headed by my second promotor Prof. B.F.A.M. van der Laan.

This thesis describes the development of technological devices that restore speech in laryngectomy patients whose larynx and vocal folds have been surgically removed due to cancer. As such, this thesis describes a technological design, an artefact. The work is scientific in that sense that the requirements to which the design should adhere, and the verification of the resulting artefact's characteristics to those requirements, have all been scientifically assessed.
The development of these devices, developed along the lines of the design methodology that my first promotor, Prof. G.J. Verkerke and I also teach, is explained in this thesis, supported by submitted or published papers, and patents.
The devices described in this thesis restore the ability to speak in laryngectomy patients. Moreover, they restore the ability to speak naturally, without thinking, intuitively. The devices thus not only restore the ability to speak, they restore part of the freedom to communicate and act, thus restoring the quality of life that the patient has lost, although declared medically cured.

1

Introduction and problem definition

Introduction and problem definition

In Europe over 40.000 people per year are diagnosed with some form of cancer of the larynx.[2] And although smoking and drinking are of major influence in the probability of acquiring laryngeal cancer, no one is safe. In the USA in 2007 about 32% of that patient group will undergo total laryngectomy as part of their treatment.[3] The larynx is the organ in the neck that connects the throat to the trachea (windpipe) and esophagus (gullet) (fig. 1a): it connects the inside of the body to the outside world of air, food and water, and contains our vocal folds, with which we communicate and express our emotions. Laryngectomy is the surgical procedure in which the larynx is surgically removed from within the neck.

Despite advances in conservative laryngeal surgery and radiotherapy, laryngectomy still remains the most frequently performed surgical treatment for this type of cancer.

Additionally, laryngectomy is often used as a last resort if minimally invasive surgery, radiotherapy or chemo-radiation treatment have proved unsuccessful.[4] As a result of the laryngectomy, the patient now has to learn to live without his/her larynx and consequently, the vocal folds (fig. 1b). After the procedure, the trachea is sutured to the base of the neck where it forms the tracheostoma (fig. 2); patients now breathe through their stoma instead of their mouth. Since the passageway to the nose no longer exists, inhaled air is no longer pre-heated, filtered or moisturized by the nose. This causes excessive mucus forming in the trachea and a higher susceptibility to infection. Both the inability to speak and the lack of air pre-conditioning obviously reduce the quality of the patients' lives,[5] and although these patients are now (often) free of cancer, their lives have dramatically changed.

Rehabilitation devices

Fortunately, many devices have been developed to improve this situation for laryngectomy patients.[6] Studies show that laryngectomy patients generally benefit from heat and moisture exchange (HME) filters to moisturize, filter and pre-heat inhaled air,[7] and patients are therefore encouraged to use these filters over their stomas.

Absence of speech is the second disability after laryngectomy, and often found to be the most severe.[8] Restoration of speech can generally be achieved in 3 different ways. The first way is to use an external sound source that can introduce an audible tone into the mouth that the patient can articulate into speech. A familiar example is the electrolarynx; an electric shaver-shaped vibrator that is pressed against the floor of the mouth to create a humming tone inside the mouth. Good speech with an electrolarynx is typically difficult to learn, but the advantage is that instantaneous, although weak, speech is possible.

The second, and oldest, speech method for the laryngectomy patient is injection speech; the patient swallows a gasp of air, and upon releasing (burping), the consequent sound can be articulated into speech. Short sentences are thus possible with reasonable, but varying, results. Only about 30% of the laryngectomy patients are able to master this technique.[9]

The third way, and currently the golden standard in the western world, is tracheo-esophageal speech. This method requires the insertion of a small check valve in a surgically punctured opening between the esophagus and trachea (figs. 1b, c, d). The shunt valve is inserted during the laryngectomy procedure. After inhaling, the patient can close the stoma with a finger,

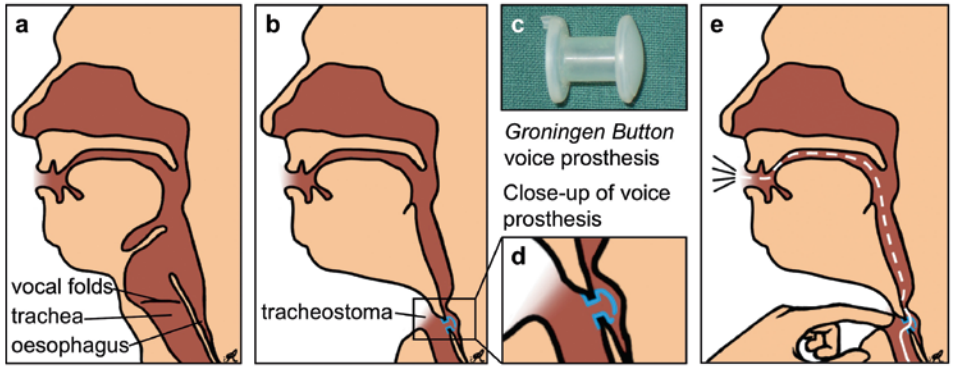


Figure 1. a) A normal larynx in a cross-sectional view of the head. b) a cross-sectional view of a laryngectomy patient's head with the larynx removed and a shunt prosthesis (see figure c and close-up d) placed between the trachea (windpipe) and esophagus (gullet). The patient can speak by rerouting inhaled air through the shunt valve at closure of the tracheostoma (e). The rerouted air will cause the esophagus to vibrate, the sound of which the patient can articulate into speech.

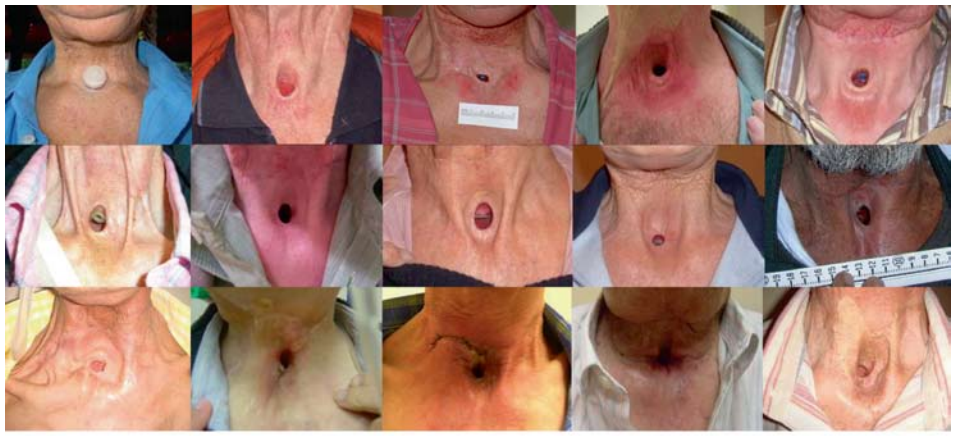


Figure 2. Photographs of a random selection of stomas taken from the peristoma geometry study described in Chapters 2 and 3.

after which exhaled air is rerouted via the shunt valve (fig. 1e). The esophagus will now vibrate as with injection speech, and allows the patients to articulate speech. It is this last method of speech restoration that is the topic of this thesis.

Since patients are advised to use HME filters on their tracheostoma, many filters are combined with a valve for manual closure of the stoma, allowing patients to use tracheo-esophageal speech.

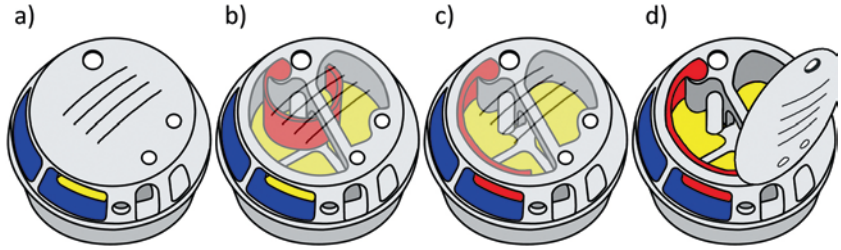


Figure 3. a,b) The Provox FreeHands HME in its normal position. Air flows through the HME (yellow) and the breathing holes. b) With normal breathing, the membrane (red) does not move. c) On strong exhalation, the flexible membrane is pushed away to block the breathing holes to allow speech. d) With even stronger exhalation (coughing), the cough-relief valve opens.

To avoid closing the tracheostoma with a finger, automatic tracheostoma speech valves (ASVs) have been developed. The ASVs can close without using fingers or hands and can be hidden out of sight by a scarf or collar. They have been very well received by patients, and are known to improve the overall quality of life.[9,10]ASVs have already been on the market for some time.[6,11,12] The most commonly used ASVs are the Atos Provox FreeHands (Atos Medical AG, Hörby, Sweden) adjustable tracheostoma valve (see fig. 3 for its workings), the Blom-Singer ATSV standard and ATSV II (Blom-Singer adjustable Tracheostoma valve, Helix Medical, Carpinteria, USA) and the Adeva Window (Adeva Medical, Lübeck, Germany). For a systematic overview of available tracheostoma valves, see Ten Hallers et al.[6]

All commercially available ASVs close upon strong exhalation by the patient, remain closed while the patient speaks, and open again when the pressure drops. This method unfortunately wastes valuable air during closure and, as pressure should be maintained during speech, this makes long sentences with pauses between words impossible.
A new ASV principle based upon inhalation has been demonstrated previously.[13–15] This principle eliminated problems associated with automatic speech valves based upon the principle of closure upon exhalation. Major improvements of this mechanism, compared to exhalation valves, include the possibility of prolonged speech, the possibility of coughing without removing the valve, and the prevention of wasting exhaled air in closing the valve.[14] Yet, never before has this principle been made into a commercially viable product.

Tracheostoma patches

All these devices – heat and moisture exchange filters, manual speech valves and automatic speech valves – need a durable interface with the tracheostoma.
Generally, two widely accepted fixation methods are available for the attachment of speech valves and HME filters to the stoma: the cannula-like intra-tracheal devices or, the less invasive, adhesive stoma patches.
Bień et al.,[16] reported that 70% of the patients use a stoma patch with a HME filter attached to it. In a single study of Hilgers et al.,[11] 78% of the patients use stoma patches, but only 25%

use an ASV on a daily basis. Lorenz et al.[9] shows even lower ASV use (< 20%), and Op de Coul et al.[10] reports only 15% daily compliance to a new type of ASV, whereas they both show that the benefits of handsfree speech are numerous.
Automatic speech valves exert considerably more stress on the interfacing agent (form fit in case of intratracheal devices or glue in case of adhesive patches) than the manually closed HMEs, because in the manually closed valves, pneumatic (speaking) pressure is countered by the finger that keeps the valve closed. This means that ASVs pose much higher demands on the quality of the stoma interface.
And this is exactly what Ten Hallers et al.,[6] Hilgers et al.[11] and Op de Coul et al.[10] show: poor fit, poor gluing capability and skin irritation are the main reasons for not wearing ASVs and their adhesive stoma patches. Currently, the only way to address the problem of poor fit is by casting a tailor-made ASV interface, a laborious and expensive solution. Or one could use several adhesive patches per day or, of course, not use automatic speech valves at all.[17,18]
Why do existing adhesive patches adhere so poorly to the patient's stoma?
The stomas of most patients are situated in a deep recession in the neck, with the area around the stoma, the peristoma, forming a concave shape enclosed laterally by the sternocleidomastoid (SCM) muscles, cranially by the chin and caudally by the chest. This recession is considerably deeper than depicted on most schematic drawings in literature and commercial communication.
Most available adhesive patches however, are completely flat, the most concave types (the Blom-Singer Tracheostoma Valve Housings BE6038 and BE6039, Helix Medical, Carpinteria, USA and the Atos XtraBase, Atos Medical, Sweden) being only around 7 mm deep. A quick field survey amongst Dutch ENT surgeons showed that stomas are often deeper than 7 mm. (fig. 2 shows a random selection of stomas; only a few of which can be classified as flat). Therefore we felt there was a mismatch between the patches available and present stoma anatomy, especially in the assumed stoma depth of the patient. In literature, however, there was not sufficient data on the post-surgical anatomy available to improve patch design. Chapters 2 and 3 extensively describe the study in which we measured and investigated shapes and sizes of tracheostomas in patients and the consequent problems these patients have with applying and wearing stoma patches.
Another reason for this mismatch of patch and stoma is the use of relatively stiff materials in the patch. Most materials used in not only the patch but also the valves are hard, stiff materials that do not match the softness and flexibility of the tissue they interact with: the patient's skin and musculature and the clothing materials surrounding the stoma.

The design method

This thesis describes the development of the iValve, automatic speech valve and the iPatch, a contoured adhesive stoma patch. The development of these devices was guided by an engineering scheme, a method of designing. This method, a method we also teach to our biomedical engineering students (based upon the work of Eger et al.,[19] Roozenburg and Eekels,[20] Van den Kroonenberg and Siers[21] and Buijs and Valkenburg[22]) provides a means to structure the design process in such a way that the chance of overlooking a viable solution is minimized and that developing an optimal solution to a problem is maximized.

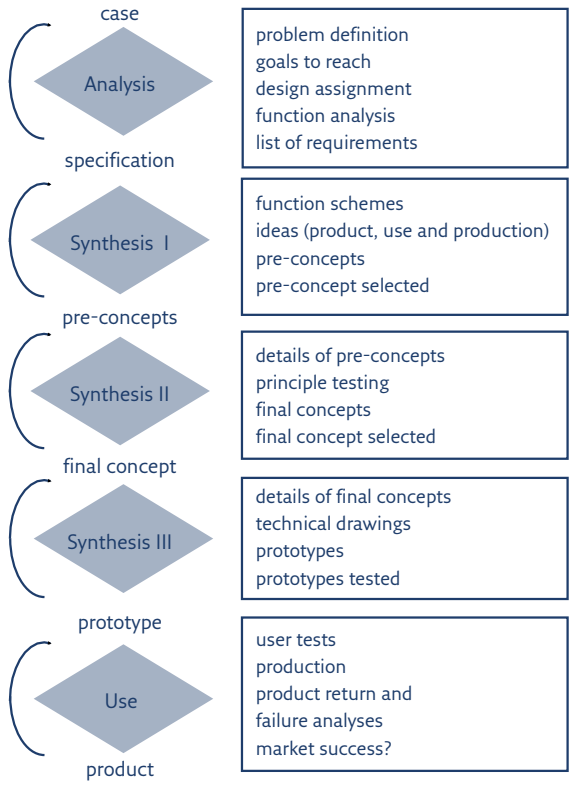


Figure 4. The design method used at Biomedical Engineering.[20]

Figure 4 shows the structure of the design method: starting with a case and ending with a product in use. Each phase in the method consists of an alternating period of information expansion and consequent information selection and processing. Thus each phase is depicted by a diamond shape representing the consequent expansion and reduction of information. After each phase, the end result of the phase is checked against the goals set for that particular phase (depicted by the returning arrows). If the end result is not considered adequate enough to reach the set goals, the phase will have to be repeated: the design method is an iterative method and often some phase is repeated either in total or on a different (design detail) scale.

Thus the method not only gives structure to the design process, it also incorporates "quality checks" along the way.

The third feature of this method is that the analysis phase and synthesis phase have been strictly separated: only if sufficient information about any problem within a case has been thoroughly analyzed and reported can the developer start to conceive means to reach the goals, on the basis of the problem definition. All these features ensure that the developed

product will solve the actual experienced problem and increase the chance of a successful innovation. It will become clear from the rest of this thesis that this product is not a tangible device per se, but could relate to any solution. The mechanical devices of this thesis are selected from a range of viable solutions that could be pharmaceutical, economical, psychological or even philosophical in nature.

Analysis phase: problem analysis

Key factor in a good development trajectory is problem definition: an encyclopedia-like lemma describing the problem, problem owners and problem context (in this case Sections 1.1 and 1.2 of this thesis). In this thesis there are six problems concerning voice restoration and rehabilitation devices which became clear from the problem definition:

1. All commercially available ASVs close upon strong exhalation by the patient, remain closed while the patient speaks, and open again when the pressure drops. This method wastes air during closure and makes long sentences with pauses between words impossible.
2. The patients' use of stoma patches, filters and automatic speech valves is generally very low, whereas the benefits of filters and handsfree speech are numerous.
3. Current automatic speech valves exert considerably more stress on the interfacing agent (form fit in case of intratracheal devices or glue in case of adhesive patches) than manually closed filters.
4. Poor fit, poor gluing capability and skin irritation are the main reasons for not wearing automatic speech valves and their adhesive stoma patches.
5. Most stomas are considerably deeper than depicted in literature, while most available adhesive patches are completely flat: there is a mismatch between available patches and stoma anatomy.
6. By far most medical devices, implanted or on the body, are made out of hard, stiff materials. Since many of these devices are surrounded by soft biological tissue, there is often a mismatch in stiffness.

Goals

From these selected problems, the goals for the end product are derived.

First, there is a primary goal. This goal is crucial to any innovative project: if this bottom line is not reached – in this case, if the patient's quality of life is not increased – the product cannot be regarded as successful. Often the primary goal is part of the mission statement of the executing company, because it is so important:

To increase the patients' quality of life.

This primary goal has several sub-goals and these are the working goals for the project. But although it is so obvious: never should the primary goal be forgotten, or the project will not be successful.

1. To enable speaking in longer sentences with mid-sentence pauses.
2. To increase the use of automatic speech valves and filters in particular, and stoma patches in general, by laryngectomy patients.
3. To reduce the stress on stoma-interfacing devices.

4. To realize a better fit and adherence of stoma-interfacing devices to the stoma.
5. To reduce the geometrical and stiffness mismatch between stoma and stoma-interfacing devices.
6. To match the stiffness and hardness of devices to the surrounding tissues.

Design assignment

Next, we formulate the design assignment. This is also where we choose to use a mechanical device to realize our goals. This means that, although we could recognize that a change of reimbursement systems, or the introduction of surgical techniques would help reach our goal, we distinctly opt to develop (a) mechanical device(s). The design assignment derived from the goals, and underlying this thesis is:

To realize (a) mechanical device(s) that allows laryngectomy patients, who can currently speak using manually occluded speech valves, to speak handsfree, better than with the widely used commercially available automatic speech valves (e.g., the Atos Provox FreeHands). "Better" means exceeding the Atos Provox FreeHands in attachment time of the valve to the stoma, ease of speaking, ease of controlling, general preference of the patient and overall economy. This will be achieved by developing a new type of automatic speech valve that works on the principle of inhalation as described by Geertsema et al.[15] This means that the valve should switch to speech mode at strong inhalation and switch back at strong exhalation, and facilitate speech in speech mode. The valve allows inhalation at all times. The valve could be a one-day disposable device or a non-disposable device. It should have a much lower manufacturing cost, and much lower total cost of ownership than the ATOS FreeHands, making several business models possible. The Valve should incorporate an HME filter, or combine with existing HME filters. The valve should preferably come equipped or supplied with a stoma-interfacing means that provides comfortable and functional interfacing of the valve to the stoma at least one entire day and should thus possess an interfacing quality far exceeding commercially used adhesive patches.

Both valve and patch will be made out of rubber-like materials to have a better match with the stiffness of the surrounding biological soft tissues and to allow for function integration lowering the overall product cost as compared to the commercially available valves.

Function analysis and functional components

In order not to overlook any solutions, a meta-analysis can be performed in which the product to be developed is made as abstract as possible. This can be done according to the MEI method (Matter, Energy, Information: see fig. 5) as in figure 6.[20] The advantage of such abstractification is that these blocks can be moved around and swapped in order to investigate problems, possible strategies and principle solutions, without the developer being inhibited by thoughts about practical implications or traditional views.

Less abstract schemes such as figure 7 can also be of great help in always having a clear overview of the functional requirements of the product. Figure 7 is a fundamental design, a design of principle (and as such usually a product of the synthesis phase). Figure 7 is one of many simple abstractions of the basic valve functions that can be used as a "summary of functions". For instance, the steady states of 'speaking' and 'breathing' mode are maintained here by magnets; but could have any embodiment in the end product.

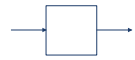
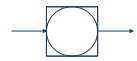
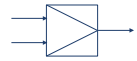
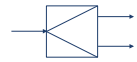
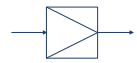
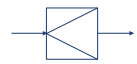
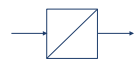
Name/function	verbal meaning	symbol
Transport	Transport, conduct, move, pump, relay	
Store	Store, keep, hold, memorize	
Connect (and separate)	Add, print, mix, connect, stir (cut, distill, scrape, read, saw, distribute)	 
Transform	Flatten , grind, parse, translate, step-down, adapt	 
Convert	Drive, power, time, use, control, generate, convert, burn	

Figure 5. Function schemes: building blocks that can be used to abstractify any system.[20]

An electrical analogon scheme could be used as well to show required airflows and paths. These are merely tools to represent the assignment in such way that the designer maintains an overview during the process.

The building blocks and product functions from the diagrams of figure 6 and 7 can be translated into product functional components that describe the parts and mechanisms required in the automatic speech valve design.

1. HME and filter (foam or fabric or any other filtering material)
2. Activator mechanism(s): Controlling when the valve switches between speaking and breathing
3. Locking mechanism(s): Maintaining speaking or breathing position until change required
4. Switching mechanism(s): Blocking (speaking mode) or unblocking (breathing mode) exhaled air
5. Inhalation provision (providing inhalation at all times)
6. Stoma interfacing (interfacing to standard adhesive patches, directly to stoma ...)

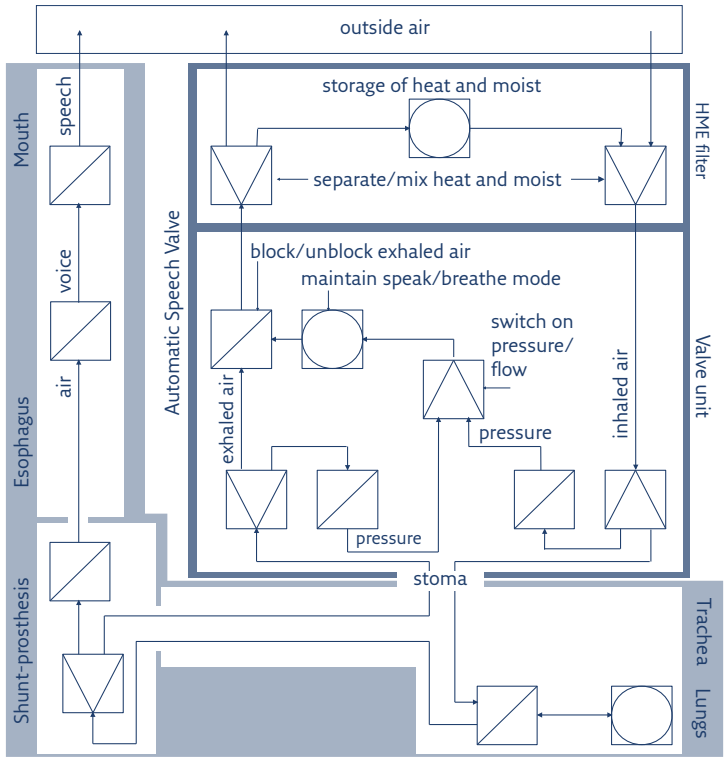


Figure 6. Function scheme of patient and tracheostoma valve according to Geertsema's inhalation principle. This function scheme comprises all the required functions according to the building blocks of figure 6: these blocks can be moved around and swapped in order to investigate possible strategies and principle solutions.

List of requirements and wishes

The analysis phase is concluded with a list of requirements and wishes, which is a list of specifications to which the product to be designed must adhere (the requirements) or should preferably accommodate as much as possible (with regard to the wishes). This list specifically does not contain descriptions of *how* this should be achieved, but only how the product should ultimately perform. This way of formulating requirements is called *functional specification*.

The items in the list must be formulated in an unambiguous and fully operational way. A good list of requirements and wishes is formulated such that even outsiders – should they design a product according to the list – must produce an optimal product. The list of requirements that was used throughout the development of both the iValve and iPatch has been detailed throughout the project using the numerous studies described in the following chapters. Without these numerical specifications, the list looks like this.

Requirements of the iValve:

- The device facilitates handsfree speech in laryngectomy patients who can already use tracheo-oesophageal speech.
- The device allows pausing mid-sentence without loss of phonation air.
- The device provides more air for phonation than commercially available handsfree valves.
- The device allows laryngectomy patients to breathe comfortably.
    - The device's airflow resistance in breathing mode is comparable to or lower than that of commercially available handsfree valves and comparable to the airflow resistance of the healthy airway.
- The device is as easy or easier to operate than commercially available hands-free valves.
- The device allows speaking that is easier than or similar that allowed by commercially available handsfree valves.
- At equal phonatory sound pressure levels, the device generates lower tracheal pressures than commercially available handsfree valves do.
- The device can be used as an appendage to tracheostoma patches or tracheostoma cannulas worn on tracheostomas.
    - The device must either fit onto existing and widely used commercially available stoma patches or the device must be integrated with a stoma attachment aid.
- The device operates in two settings: a breathing setting in which the patient can comfortably breathe in and out, and a speaking setting in which the device will block exhalation.
- The device switches from breathing to speaking mode through strong inhalation exceeding normal inhalation flow.
- The device switches from speaking to breathing mode through strong exhalation exceeding normal speaking pressure.
- The device is comparable or smaller in size than commercially available handsfree valves.
- The device incorporates an HME filter that filters and moisturizes inhaled air, or can be combined with existing HME filters.
- The device withstands, i.e., will not dislodge at, the patient coughing.
- The device is as easy or easier to attach and detach than commercially available handsfree valves.
- The device is more economical to manufacture than commercially available hands-free valves.
    - The device contains fewer parts than commercially available handsfree valves.
    - The device is made for the major part out of soft and flexible materials.

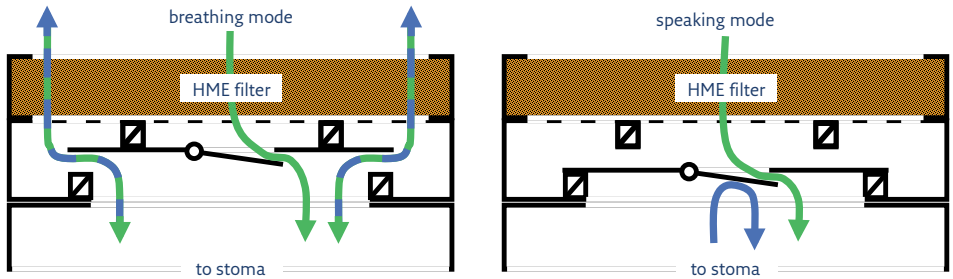


Figure 7. Higher-level abstract scheme to understand the functional requirements of the automatic speech valve. For instance, the state memory of 'speaking' and 'breathing' mode are embodied here in magnets (the crossed rectangles); but could have any embodiment in the end product.

Requirements of the iPatch:

- The device facilitates (and will not detach at) handsfree speech in laryngectomy patients.
- The device can used as an interface between the tracheostoma and widely used (commercially) available HME filters, speech valves, shower aids and our iValve.
- The device incorporates a standard connector that fits widely used (commercially) available HME filters, speech valves, shower aids and our iValve.
- The device facilitates comfortable breathing in laryngectomy patients.
- The device interfaces, and remains attached to, most tracheostomas that prove difficult to connect to standard, widely used (commercially) available stoma patches.
- With normal use, the device adheres to the stoma for at least the active part of the day.
- The skin tolerance to the device is similar or better than that to standard, widely used (commercially) available stoma patches.
- The device is as easy or easier to apply than standard, widely used (commercially) available stoma patches.
- The device is as economical or more economical to manufacture than commercially available adhesive patches.

Synthesis phase, overview

The synthesis phase is the phase in which we start thinking about schemes or designs to meet our goals. We can guide and limit these schemes and designs by the technology, skills and strategies that we have, or want to have, at our disposal. This is why we choose to develop a mechanical device and not focus on, for instance, changing reimbursement systems in countries of low patch use. Strategy also gives roadmaps so that skills and technology developed for one product can be put to use on future products for the continuity of the organization.

Figure 8. iValve v1 on a patient, attached by means of an Atos Optiderm adhesive patch.

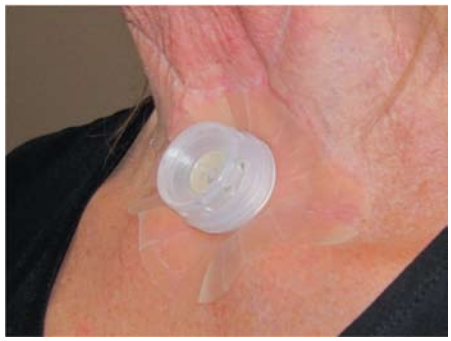


Figure 9. iValve v2 on a patient, attached by means of an Atos Optiderm adhesive patch that has been radially incised by the patient to improve the fit.

Generating solutions

The next step in the design method is the generation of ideas or solutions. The synthesis phase of the design method separates the first synthesis phase in which any idea, no matter how strange, is welcomed to the idea pool, from the second synthesis phase, in which ideas are preselected according the list of requirement and wishes. The third synthesis phase refines some selected, and all viable, ideas to a level where they can be tested. At this level ideas are called concepts, and concepts are prototypes when they are built. Tools like figure 7 were used to remind the designers of the requirements of the device throughout all synthesis phases, and every prototype is thus an embodiment of that scheme.

Chapter 4 shows the investigation into the main operational principle of this scheme, the buckling mechanism. In this chapter, a switch with two steady states is described. Checking if such a mechanism could work at all is typically done in synthesis phase 1, while extensive research on how to make it work well, is done in the second synthesis phase (see fig. 4). In synthesis phase 3 several concepts were detailed, amongst which are those of Chapters 5, 6 and 7. The design and testing of the prototype of the automatic speech valve are described in Chapters 5 and 6, and Chapter 7 shows the iValve v2 (fig. 9), the valve that was developed in

concurrence with the iValve v1 (fig. 8) but in which much of the knowledge acquired testing the v1 had already been incorporated. Chapter 7 also describes the iPatch version 1 and 2: contoured stoma patches that were designed (using the data from the stoma geometry studies of Chapters 2 and 3) to conform better to the shape of most stomas. The prototypes of Chapter 7 have already shown to be viable solutions in (limited and thus preliminary) patient tests, but full-scale clinical testing will have to confirm this.

Designing in rubber

We decided that, for a number of reasons, building our devices in soft materials would give us an innovative advantage over existing products and developments in the field of voice restoration and speech valves.

Using rubber materials would prevent the mismatch in stiffness and softness between device and tissue that we found in our analysis phase and would therefore contribute to a better biocompatibility. Using soft, elastic, materials such as rubber also opens up possibilities of elaborate function integration: the combining of many shapes required for certain functions into a single part so that the consequent device contains as few parts as possible. Reducing the number of parts reduces the number of assembly steps, which in turn reduces the assembly cost, chances of failure of connections and even reduces the effort to have the device certified by notified bodies. The knowledge of rubber and designing and rubber would also be transferable to other projects involving implants.

In addition, rubber is much more forgiving to design in: rubber parts, when made by injection molding, require less drafting in the mold, and even undercut in the mold (undercut occurs when part of the mold is so enclosed by the molded design that it is impossible to take out without breaking) is allowed to some extent. Molded parts can be stretched over the undercut or simply blown out of the mold cavity by air pressure. Designing devices in rubber requires no extra skills in comparison to designing devices in hard materials, but it does require a different way of thinking. Designing in rubber requires holistic thinking, since all features constantly interact through their elastic relations: there are no real defined dimensions to the product. Designing in rubber, using rubber to its full potential, requires a complete new set of design rules. Rubber can be stretched, rolled, inflated and flipped inside-out and outside-in, which are deformations almost impossible in hard, "traditional" materials. Using rubber to its full potential opens up whole new possibilities and applications far beyond the known applications of tubes, tires, driving belts, elastic bands and seals. It is remarkable that silicone rubber kitchen utensils are only now beginning to appear on the market, while the material itself has been around for decades at comparable, albeit somewhat higher bulk prices.

Elastomers used for implantable biomedical applications include silicone rubber, polyurethane (PUR) and a number of thermoplastic elastomers (TPEs), like SIBS30[23] or even biological rubber-like materials.[24] For elaborate descriptions of properties of medical rubber materials and especially that of silicone rubber, see Chapter 4.

Building and testing prototypes

The synthesis 3 phase requires prototypes to be built for testing. In the phases prior to this

Figure 10. A silicone rubber product (iValve v1, capseat part) with a typical mold as used for the creation of all the rubber products in this thesis (background).

Figure 11. version family of the iValve v2 from right to left; subsequent (and thus improved) versions. Especially noticeable is the decrease in size. The family continues to the right since improvement is ongoing.

phase, details have already been manufactured in order to be tested or studied (for an example see Chapter 4), and this manufacturing knowledge is later used to build the entire product in prototype. Prototypes behave and look similar to the final product, but they are not necessarily manufactured similarly. In the case of our iValve and iPatch prototypes, these are individually molded in transparent PMMA molds. These molds are optimized for manufacturing on the available machines: the CNC turning lathes of the UMCG's own Instrument Workshop, those of GME Machinefabriek (Enschede, The Netherlands) and of Veenstra-Glazenborg (Winschoten, The Netherlands). Later production models will have to be optimized

for injection molding: but prototyping proved their working and manufacturing feasibility.
We choose to use silicone rubber for our devices due to the ease of prototyping (the used silicone rubber comes in ready-made two-component injection cartridges) and the possibility to translate the material properties to a wide variety of medical grade rubber later. Molding prototypes requires a mold containing the negative shape of the prototype, a rubber material and means to solidify (cure) the liquid into the molds (fig. 10). Our two-component silicone rubber is heat-cured at 70°C in a standard convection oven.
From further research it became apparent that although translating the same mechanical functions of these devices to soft materials seemed possible, very little experience was to be found. Chapter 4 describes a study in which we investigated one particular mechanism used in the iValve designs: the bi-stable buckling mechanism.
Altogether, around twenty molds were made, producing hundreds of prototypes (for a version family of iValve v2's, see fig. 11). Eventually only a few of those were tested on patients. These tests are described in Chapter 6. After that, the product will have to be prepared for mass production and can be introduced into a small selected patient group. The real product test comes with commercialization and will provide the last feedback loop in the design process. This last loop will be touched upon in the General Discussion section (Chapter 8).

2

Stoma geometry and patch use

Does the patch fit the stoma? A study on peristoma geometry and patch use in laryngectomized patients

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Abstract

Background: Heat-and-Moisture-Exchange (HME) filters and (handsfree) speech valves contribute significantly to the quality of life of laryngectomy patients. These devices, however, are not widely used. This is mainly attributed to poor fit of the adhesive patches to the stoma site. Current patches are not based upon objective (peri)stoma geometry because this geometry is unknown.

Objectives: This study investigates the geometry of (peri)stomas in relation to patch use. The study delivers data enabling improvement of tracheostoma interfaces, specifically addressing patients currently unable to use stoma patches.
Design: Observational anthropometric study of the (peri)stoma of laryngectomy patients.

Participants: 191 Laryngectomy patients from various countries, at least 1 year post-operative.

Intervention: (Peri)stomas were photographed and measured. Patients completed a questionnaire on patch use. The concavity of commercially available patches was measured.

Results: In countries with a financial reimbursement system, 58% of the patients use patches, compared to only 9% in other countries. Patches stay *in situ* for an average of 33.3 hours. Patch and non-patch users differ on five out of ten measured geometrical parameters. The most striking differences are that patch users have much shallower peristomas (13 versus 18 mm), and stomas far more parallel to the anterior neck plane. The deepest commercially available patch is only 7mm deep.

Conclusions: This study provides detailed (peri)stoma geometry data of a divers population, and for the first time in relation to patch use. It reveals a serious mismatch between patients and patches. With these data, new patches can be developed that could dramatically improve rehabilitation after laryngectomy.

Introduction

Laryngectomy is currently the most frequently performed treatment for advanced-stage laryngeal carcinoma. The patient's initial inability to speak and the lack of air pre-conditioning have both been shown to decrease quality of life.[5]
To enable speech, finger operated Heat-and-Moisture-Exchange (HME) filters and, for hands-free speech, Automatic tracheoStoma Valves (ASVs)[6,9,11,12] have been on the market for some time, or are in development.[15] Heat and moisture exchange (HME) filters effectively improve the tracheal climate[5,25] and generally benefit laryngectomy patients.[7,16] In general, two widely accepted fixation methods are available for attachment of ASVs and HME filters to the stoma: intra-tracheal devices or adhesive stoma patches.

The use of patches, filters and valves varies. Literature suggests 78% compliance for a manually operated valve with HME filter,[26] 78% for any stoma patch in general,[11] 70% for a patch with HME filter[16] and 15-25%[9,10,16] compliance for daily use of an automatic speech valve. ASVs in particular are not widely used due to compliance issues.
ASVs exert considerably more stress on the attachment system than the manually occluded HMEs during speech. With manual occlusion, the pneumatic pressure is countered by finger pressure, while in ASVs all speech pressure is exerted on the stoma interface. Poor fit, poor gluing and skin irritation (mostly related to the use of glue) are mentioned as the main reasons for not using ASVs,[6,10,11] and are all interface related. Currently, the only way to improve this is by casting a tailor-made ASV interface,[17,18] which is a laborious and expensive solution.
If we want to increase ASV use, this stoma-interfacing needs to be improved. The unknown geometry of the stoma and peristoma, however, hinders this development. Therefore we set up a study to obtain all information required to improve stoma-interfacing. Improved adhesive patches will increase the use of these patches, their connected speech valves and HME filters, and can significantly improve the quality of life in this patient group.
The study was split up into a clinical and an anthropometrical part, published in parallel. In the clinical paper we investigated the influence of the surgical procedure on the shape and size of the (peri) stoma.[27] This paper showed that the surgical approach, where the sternocleidomastoid is incised in order to decrease the stoma depth, is of very limited effect on the depth of the stoma, and therefore on the improvement of patch fit.[27] It also showed that the number of removed tracheal rings is of no influence of the stoma depth. The clinical paper presents only the stoma geometry for the entire measured population.
This anthropometrical paper presents not only the geometry of the (peri)stoma of laryngectomy patients in general, but especially the difference in stoma geometry in patch users and non-patch users specifically. Commercially available adhesive patches were also measured and compared to the collected stoma geometry, in order to investigate a mismatch between peristoma shape of non-patch users and the commercially available patch shape. This paper is the first study ever to present this difference in peristoma geometry. With this data it is possible to develop stoma interfaces, e.g. adhesive patches, specifically for this large group of unaddressed patients, improving HME and ASV use, with all the associated increase in quality of life.

Materials and Methods

Patients

A total of 191 patients (10% female) from various institutions in 8 countries (The Netherlands (n = 84), Belgium (n = 20), France (n = 12), New Zealand (n = 5) Italy (n = 6), Spain (n = 17), Portugal (n = 26) and Egypt (n = 21) were randomly included from routine out-patient clinic visits. The average age was 65.5 years (range 35-92 years). Inclusion criteria were total laryngectomy at least 1 year prior to the study. Exclusion criteria were irritation of the (peri)stoma, tumor recurrences, or poor wound-healing in such a way that measuring the (peri)stoma was deemed unhygienic, painful or difficult. Data from all institutions were pooled to establish a homogeneous database, effectively spanning different institutes and surgeons, delivering the (peri)stomal geometry of a population as large as possible.

Ethical considerations

This study was approved by all the respective institution's ethics committees. All included patients were scheduled for routine check-up (e.g shunt valve changes) and signed informed consent forms.

Methods

To characterize those geometrical parameters of the (peri)stoma that influence stoma-interfacing, we derived 10 parameters. More parameters would only slightly improve characterization, but increase measuring time. The selected number proved a good balance in practice. All patients and investigators completed a questionnaire. Photographic images were taken of each patient's neck to enable retrospective verification of measured data. All photos, measurements and classifications were done with the patient sitting upright in a chair. Measurements took about 30 minutes per patient.

The protocol used in this study consists of three parts.

- direct measurement of the stoma and peristoma by means of a (tape) ruler (1 mm divisions) and Vernier Caliper (minimal accuracy 0.5 mm),
- 10 standardized digital pictures of the peristoma and neck,
- a detailed questionnaire from which the patient's age and gender and three questions related to patch use ("do you use stoma stickers or patches?", "are adhesive patches financially reimbursed by your health care provider?" and "how long do the stickers/patches stay?") were used in this study.

In addition, the depth (defined as the distance from inner to outer rim perpendicular to the stoma) of adhesive patches from the three most popular brands (Blom-Singer (Helix Medical, Carpinteria, USA), Atos (Atos Medical, Hörby, Sweden) and Ceredas (Ceredas, Antony, France)) were measured with a Vernier Caliper (minimal accuracy 0.5 mm).

The applied surgical techniques, patient history and their influence on patch use and (peri) stoma geometry were presented in our clinical study.[27]

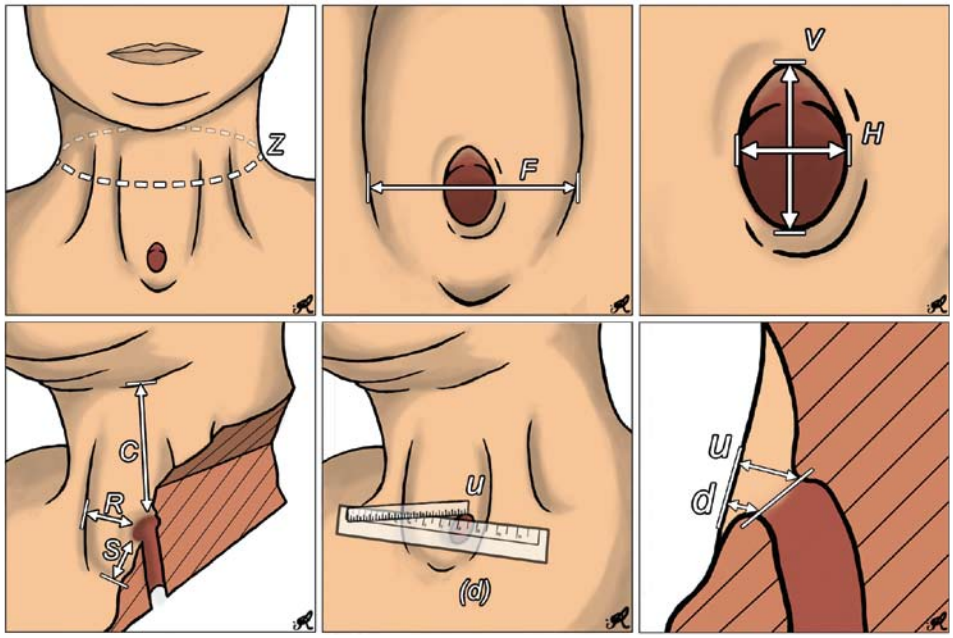

Figure 1. Parameters of the patient's (peri)stoma (distance *L* omitted in this figure for clarity).

Measured Parameters

The following ten parameters were measured according to Figure 1:

- The collar size *Z* (circumference of the neck at stoma height).
- The horizontal *H* and vertical *V* diameter of the stoma itself.

The area of uninterrupted, smooth, scar and crevice-free skin around the stoma is defined by the following parameters:

- Distance fold-to-fold *F*: the horizontal distance (at stoma centre height) between the first interruptions on each side of the stoma; in most cases these folds are the sternocleidomastoid muscles (SCM). This distance is maximized at 100 mm: distances greater than 100mm are noted as 100mm.
- Distance stoma-to-chin *C*: the distance from the stoma to the first upward interruption (often the base of the chin). This distance is maximized at 50 mm.
- Distance stoma-to-sternum *S*: the distance downward from the stoma to the first interruption, which is often the jugular notch (the recession above the sternum). This distance is maximized at 50 mm.
- Distance stoma-to-right-fold *R* and distance stoma-to-left-fold *L*: the horizontal distance between the stoma edge and the first fold proximal to the stoma (often the SCM) on the patient's right and left sides. This distance is maximized at 50 mm.
- Superior stoma depth *u*: the distance from the most superior edge of the stoma to

the plane over the first interruptions lateral to the stoma (often the depth of the deepest stoma edge seen from the plane over the SCM).

- Inferior stoma depth *d*: the distance from the most inferior edge of the stoma to the plane over the first interruptions lateral to the stoma (often the depth of the most shallow stoma edge measured from the plane over the SCM).

Calculated parameters

Four parameters are derived from the measured parameters: the ellipticity of the stoma ε(*H/V*), the average stoma depth D_{ave} ((*u*+*d*)/2), the area of the stoma Astoma (*H*x*V*x¼xπ) and the inclination of the stoma relative to the proximal folds α, in degrees (arcsine((*u*–*d*)/*V*). A higher α designates a stoma leaning relatively backward and an α of 0 designates a stoma parallel to the proximal protrusions of the SCM or proximal folds.

Statistical analysis

All parameters were analyzed using SPSS version 16.0. Statistical analysis was mainly based on descriptive analysis. The normality of all parameters was checked visually using histograms and Q-Q plots and, where needed, the parameters were log10-transformed to achieve normality. In case of normality, equality of means of parameters was tested using the independent sample t-tests. In case of non-normality, parameters were tested using a Mann-Whitney non-parametric rank test. Relations between nominal parameters were investigated using the Pearson Chi-square test. A two-tailed p-value of <0.05 was taken to indicate statistical significance.

All parameters were checked for extreme values (checked on the respective photos). Extremes were excluded from analysis. To assess the intra-observer variability, half of the patient population was measured twice by the same observer (see table 4). The intra-observer agreement was calculated using kappa coefficients. Values for kappa between 0.61 and 0.80 denote substantial agreement and values higher than 0.80 denote almost perfect intra-observer agreement.

Results

General results

In all patients (n=191), only 36% report the use of patches (table 1). In four countries (Spain, Egypt, Italy and Portugal), patches are not financially reimbursed by health insurance. With these countries excluded, 58% of the patients (n=121) report to use patches (table 1). In the group of daily adhesive patch users (all countries included, n=56), the mean patch-*in-situ* duration was 29.5 hours (33.3 hours for patients in reimbursed countries alone). Not all parameters were consistently available for all patients due to missing data and removed extreme values.

Normality of measured parameters

All parameters were checked on normal distribution. Four parameters (*C*, *d*, *R* and *L*) were not

Table 1. Patch use in reimbursing and non-reimbursing countries.

	n	% patch users in total	% patch users in responders	% non responders
Reimbursing countries*	121	41	58	28
Non-reimbursing countries†	70	9	9	1
Total	191	29	36	18

** The Netherlands (n = 84), Belgium (n = 20), France (n = 12) and New Zealand (n = 5)*
† Italy (n = 6), Spain (n = 17), Portugal (n = 26) and Egypt (n = 21)

normally distributed. The parameters *R* and *L* were normally distributed after $\log^{10}$-transformation. Descriptive statistics for the entire population are presented in table 2.

(Peri)stoma geometry and adhesive patch use
The geometry in relation to patch use was only investigated in patients from reimbursing countries that responded to the question about patch use (n=84). Five parameters show a difference in geometry (table 3) between patch users and non-patch users.
On average, patch users have thicker necks, a larger fold-to-fold *F* distance, shallower and flatter peristomas (more parallel to the front of the neck). The α in patch users was 30 degrees, compared to 91 degrees for non-patch users (P95 respectively 27 and 57 degrees inclination, table 3).

Adhesive patch depth
An investigation of the three most popular brands on the market showed that most tracheostoma adhesive patches are flat. The only concave types (The Blom-Singer BE6038 and BE6039, Helix Medical, USA and the Atos XtraBase, Atos Medical, Sweden) are around 7 mm deep.

Assessment of intra-observer reliability
Calculated kappa coefficients (table 4) show almost perfect intra-observer agreement for *V* and *C* (kappa values of respectively 0.875 and 0.881). Other measurements show substantial agreement (0.630 to 0.783) (see table 4).

Table 2. Geometry for the total population (all sizes in mm).

	Valid n	Mean (SD)	Min.	Max.	P5	P95
Circumference of the neck *Z*	166	367 (33)	280	440	310	420
Horizontal stoma diameter *H*	186	13 (4)	3	23	8	20
Vertical stoma diameter *V*	188	17 (5)	6	33	10	25
Distance fold-to-fold *F*	187	38 (16)	10	100	14	70
Distance stoma-to-sternum *S*	188	24 (9)	3	50	10	40
Superior stoma depth *u*	175	13 (9)	0	50	0	32
Mean stoma depth D_{ave}	175	13 (8)	0	35	1	28
Area of the stoma (mm^2) $A_{stoma}=(HxVx¼\pi)$	186	186 (88)	20	506	71	346
Ellipticity of the stoma $\varepsilon=(H/V)$	186	0.8 (0.2)	0.3	1.5	0.5	1.1
Stoma inclination α (degrees)	166	3 (22)	-91	91	-32	37
	Valid n	Median (Var.)	Min.	Max.	P5	P95
Distance stoma-to-chin *C**	186	40 (240)	4	50	10	114
Distance stoma to patients' right fold *R**	184	17 (87)	3	82	7	45
Distance stoma to patients' left fold *L**	185	17 (88)	5	52	8	40
Inferior stoma depth *d**	175	11 (65)	0	35	0	27

** not normally distributed (median and variance presented). Abbreviations: P5, 5th percentile; P95, 95th percentile*

Table 3. Significantly different geometry for patch users and non-patch users (Portugal, Egypt, Spain, and Italy excluded). All sizes in mm.

Patients that report patch use	Valid n	Mean (SD)	Min.	Max.	P5	P95
Circumference of the neck *Z*	37	358 (30)	290	435	290	430
Distance fold-to-fold *F*	49	44 (23)	10	100	11	100
Superior stoma depth *u*	49	13 (10)	0	50	0	26
Mean stoma depth D_{ave}	49	13 (7)	0	35	0	24
Stoma inclination α (degrees)	45	1 (24)	-91	30	-45	27
Patients that report no patch use	**Valid n**	**Mean (SD)**	**Min.**	**Max.**	**P5**	**P95**
Circumference of the neck *Z*	30	381 (24)	340	420	340	415
Distance fold-to-fold *F*	37	34 (12)	10	60	11	55
Superior stoma depth *u*	37	21 (11)	0	40	2	37
Mean stoma depth D_{ave}	37	18 (9)	0	34	2	34
Stoma inclination α (degrees)	33	20 (24)	-15	91	-15	57
Total responding patients	**Valid n**	**Mean (SD)**	**Min.**	**Max.**	**P5**	**P95**
Circumference of the neck *Z*	67	369 (30)	290	435	330	415
Distance fold-to-fold *F*	86	40 (20)	10	100	11	76
Superior stoma depth *u*	86	17 (11)	0	50	0	35
Mean stoma depth D_{ave}	86	15 (9)	0	35	0	29
Stoma inclination α (degrees)	78	9 (26)	-91	91	-38	51

Abbreviations: P5, 5th percentile;P95, 95th percentile

Table 4. Intraobserver reliability assessed by kappa coefficient.

Parameter	Kappa coefficient (valid cases)
Circumference of the neck *Z*	0.630 (94)
Horizontal stoma diameter *H*	0.698 (92)
Vertical stoma diameter *V*	0.875 (86)
Distance fold-to-fold *F*	0.783 (90)
Distance stoma-to-chin *C*	0.881 (92)
Distance stoma-to-sternum *S*	0.689 (93)
Distance stoma-to-patients' right fold *R*	0.754 (90)
Distance stoma-to-patients' left fold *L*	0.751 (90)
Superior stoma depth *u*	0.735 (92)
Inferior stoma depth *d*	0.725 (91)

Discussion

Comparison with other studies

Dirven *et al.* (2008)[28] measured (peri)stomas in a small (n=20) Dutch population and found similar stoma sizes and stoma depths: *H*, *V*, *L* and *R* respectively 14.7, 19, 15.5 and 16 mm (Dirven 2008)[28] versus 13, 17, 17 and 17 mm (this study).

The patient compliance with tracheostoma devices (measured as patch use) in this study (58%) is considerably lower than the values found in literature (78% and 70%).[11,16,26] Apart from differences in (peri)stoma geometry, the personal preferences of the practitioners, culture and climate might also have effects on patch use. The geometrical differences found in this paper are highly significant ($p \leq 0.016$) and clinically relevant.

Key findings

The major factor determining patch-use is the financial reimbursement of tracheostoma devices (58% compliance vs. 9% in non-reimbursing countries). Thus a major improvement in patch use could be obtained with less expensive patches (and connected devices) or (improved) financial reimbursement systems. No significant difference in gender amongst patch users was found, though the low number of female patients in this group (n=12) rendered fair comment impossible.

We also investigated the influence of wrinkles and folds (scars) and found these factors to have no significant influence on overall patch use. No relationship between patch use, mucus/phlegm production, *in-situ* patch duration and average stoma depth was found.

Table 3 shows that the peristoma in patch users are considerably shallower (D_{ave} = 13 mm) than the peristoma in non-patch users (D_{ave} = 18 mm). This can be largely attributed to the superior stoma depth *u* and less to the Inferior stoma depth *d*. This suggests a change in the stoma plane angle: stomas of patch users are considerably more parallel to the front of the neck than those of non-patch users. Thus the peristoma overall depth and relative stoma angle determine the quality of patch accommodation.
Considering the mean stoma depth of 13 mm in the total patient population with the 7 mm of the commercially available deepest stoma patches, the high number of interface problems in literature can be fully explained by our presented data.[6,9,11] This clearly indicates that adjusting the patch size and shape to the (peri)stoma geometry is very likely to benefit compliance and successful use of these devices.
The measurements also show that the overall space around the stoma is important to patch accommodation. In patch users, the distance *F* is larger than that in non-patch users. Yet the distances *L* and *R*, or the distance *C* or *S*, show no significant differences. Thus the area of the peristoma in non-patch users is smaller and more concave. Given this, the area for glue retention decreases. With *L*, *R*, *C* and *S* still showing ample space for a more concave patch, developing a patch that makes use of this space is clearly possible.

Clinical and practical implications
We found that non-patch users have deeper and more inclined stomas than patch users. Therefore, either the patch should be altered to fit these peristomas or an effort should be made to decrease the depth of the peristoma and the inclination of the stoma during surgery. The latter, by means of incision of the SCM, is being done in some institutions. However, as we have shown in our clinical paper, the use of this procedure has only very limited effect.[27] Alteration of the patch design should be preferred over alteration of the patient's anatomy and should at least be pursued before considering the drastic alternative. Two examples on how to improve the design of an adhesive patch for better fit are supplied in the Appendix, section 1: "How to use the presented data in stoma patch design: Two examples.".

Intra-observer reliability
The presented parameters are measurements of soft tissue, on which landmarks for reproducible measurements have intrinsic limitations. Despite these challenges, the intra-observer reliability showed high levels of agreement between the different measurements.

Conclusions

This study shows that there is a wide variation in (peri)stoma anatomy, a wide variability in the use of stoma patches, and a difference between the (peri)stomal anatomy and shape of current adhesive patches.
Patch use seems to be associated with stoma depth and with stoma inclination. Patch users have relatively shallow, parallel stomas (parallel to the frontal structures of the neck), while non-patch users have deeper, more inclined stomas (tilted backward). Patch users have on

average thicker necks and a larger fold-to-fold distance. The distances *R* and *L* do not differ between patch users and non-patch users, nor do they differ between different stoma depths.

This study of a broad laryngectomy patient population is the first to clearly demonstrate the different peristoma geometries for patients who can, and patients who cannot use patches, and to deliver the geometry to improve this situation. Deeper patches, adjusted to the geometry of patients currently without adhesive patches, are a simple and much needed step to increase patch use. This is a step that can dramatically increase the use of patches, their connected speech valves, and HME filters, and by doing so will significantly improve the quality of life in this patient group.

3

Stoma geometry and clinical implications

Worldwide, Multicenter Study of Peristomal Geometry and Morphology in Laryngectomees and its Clinical Effects.

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Abstract

Background: The purpose of this study was to improve attachment of automatic tracheostoma valves, the knowledge on tracheostoma geometry, and the preferred clinical influences. This article investigates whether the number of removed trachea rings, incision of the sternocleidomastoid muscles, neck dissection, reconstruction, time after operation, and age have any effect on the (peri)stomal geometry of the patient.

Methods: (Peri)stomas of 191 patients from 10 institutes worldwide were photographed, measured and compared.

Results: Paired comparisons between the number of trachea rings removed showed significant differences in horizontal and vertical trachea-opening diameters, but failed to demonstrate an effect in the depth of the stoma. T-tests did not demonstrate significant differences in peristomal geometry between the sternocleidomastoid-cleaved and non-cleaved group.

Conclusion: The number of removed tracheal rings during laryngectomy does not seem to influence stoma depth. However, this study does not demonstrate that cleaving the sternocleidomastoid muscle (SCM) at the time of a laryngectomy will result in a geometrically flatter stoma.

Introduction

Total laryngectomy, as a primary treatment option, is performed in patients with advanced supraglottic T3/T4 cancers, T4a glottic cancers, T3/T4 subglottic cancers, and in patients with poor general conditions or those unable to tolerate the potential respiratory complications of partial surgery.[29] Furthermore, laryngectomy is often used as salvage therapy for tumor recurrence after surgery, radiotherapy, or chemo-radiation treatment.[4] Total laryngectomy implies a permanent tracheostoma and laryngeal voice loss with permanent separation of the upper respiratory and digestive tracts. Patients who lose their larynx due to total laryngectomy can regain voice and speech in several ways.
The best quality of voice and speech is obtained using a shunt valve also known as tracheoesophageal voice and speech.[30-34] Furthermore, voice and speech restoration by tracheoesophageal speech has a higher success rate than restoration by means of esophageal speech.[35] With the introduction of low-pressure shunt valves,[36-38] additional myotomy of the cricopharyngeal and lower pharyngeal constrictor muscles have improved these results even further.[39-41]

The shunt valve is a one-way valve that operates as an artificial fistula connecting the trachea and esophagus. When the tracheostoma is occluded, the shunt valve allows air to be forced in the direction of the esophagus. This air induces vibration of the pharyngoesophageal mucosa and thus aids voice production and speech. However, this means of speech requires the patient to use his/her hand to occlude the tracheostoma. As a result, the patient needs at least 1 hand to initiate speech, which restricts mobility, points to the disability, and is deemed unhygienic. To tackle this problem, the first automatic tracheostoma valve was applied in 1982, which reduced the need for digital closure of the tracheostoma.[12,42] This type of automatic tracheostoma valve, which is open throughout normal breathing, relies on a short burst of exhaled air when the patient wants to initiate speech. This short burst of air causes closure of the automatic tracheostoma valve. During inhalation, the automatic tracheostoma valve opens again. A recognized difficulty in automatic tracheostoma valve use is its attachment to the tracheostoma. Adhesive patches are applied to attach the automatic tracheostoma valve to the soft tissue surrounding the stoma (peristomal area). The aim is to produce an airtight seal. The length of time of this sealed fixation method is very inconsistent. Factors influencing the duration of this seal are back pressure, phlegm production, and peristomal anatomy.[42] As a short burst of air is needed to close the current automatic tracheostoma valves, its use is considered to exert substantial stress on the interface between the stoma and the valve. Ten Hallers et al.[6] and Hilgers et al.[10,11] show poor fit, poor gluing capability and skin irritation of the adhesive patches as the main reasons for not wearing ATVs and adhesive stoma patches. The problem of fixation can be dealt with by applying a tailor-made ATV interface.[17,18] This is done at considerable cost and, therefore, is not feasible for all patients.

To improve the automatic tracheostoma valve interface, knowledge of tracheostoma geometry and its clinical influences is imperative.[12] We studied the geometry of the peristomal area as it is changed by laryngectomy and we investigated the effects of several factors on

the peristomal geometry. The factors we studied included the number of removed trachea rings, the incision of the sternal attachment of the sternocleidomastoid muscles (SCM), neck dissection, reconstruction, time after operation, admission of post-operative radiotherapy, and age of the patients.

To our knowledge this is the first paper to present data of clinical implications on peristomal geometry in laryngectomy patients.

Material and Methods

A total of 191 patients from 10 institutes across the world were included in this study. Inclusion criteria were patients who had undergone a total laryngectomy at least 1 year previously. Exclusion criteria were infection, ulceration or edema of the stoma, tumor recurrences, or poor wound-healing in such a way that measuring the stoma was deemed unhygienic, painful or difficult. All included patients were scheduled for routine check-ups (e.g., shunt valve changes). This study was approved by the local medical ethical committee. The study protocol consisted of 4 main items: (1) imaging of the stoma, (2) 10 lengthwise measurements of the stoma and peristomal region, (3) classification of the stoma in predefined categories, and (4) a structured study-specific questionnaire.
Measurements included the following 10 items (fig. 1):

Z circumference of the neck at stoma height,
H horizontal diameter of the trachea opening in the skin (stoma),
V vertical diameter of the trachea opening in the stoma,
F distance between the SCM or skin folds,
C upper border of the stoma until the hyoid or first skin fold,
S lower border of the stoma until jugular notch of the sternum,
L left border of stoma until SCM or skin fold,
R right border of stoma until SCM or skin fold,
u depth of the upper border of the stoma opening,
d depth of the lower border of the stoma opening.

Item 4 consisted of a structured study-specific questionnaire comprising 23 items regarding stoma care and stoma fixated aids. From this list, we used the following items in this study: does the patient use a stoma aid on a regular basis, and if so, how long does the stoma aid stay *in situ*? Furthermore, 27 items regarding the patient's surgical history were filled in by the examiners. The items used in this study are listed in Table 1. Finally, the date of laryngectomy and age of the patient were noted.

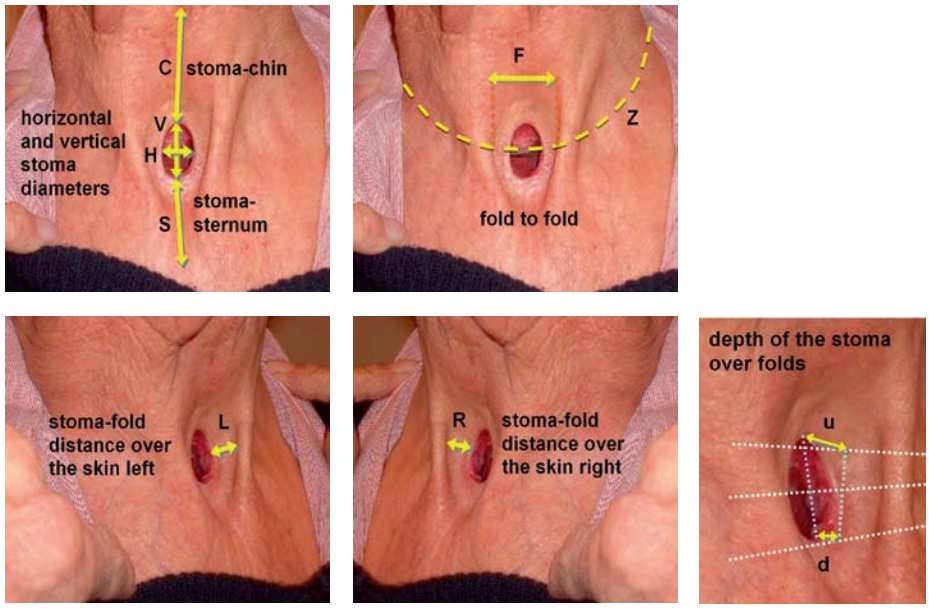

Figure 1. Visual overview of the 10 measurements on the peristomal area.

Table 1. Overview of the data studied.

Factor	Variable
No. of removed trachea 'rings'	1,2,3,4 rings
SCM incision	Yes/no/not applicable
Age	Age in months
Neck dissection	None/one sided/bilateral
Type of neck dissection	None/radical/modified/selective
Pharynx reconstruction	Yes/no
Radiotherapy	Yes/no

Abbreviation: SCM, sternocleidomastoid muscle

Statistical Analysis

Effect of Number of Removed Trachea Cartilage ''Rings'' during Laryngectomy
The total cohort consisted of 3 groups having 2 (n = 51), 3 (n = 36), and 4 (n = 10) tracheal rings removed, respectively. As only 2 patients had 1 trachea ring removed, these cases were excluded from further analysis. The 3 groups were compared using analysis of variance (ANOVA). In case of significant results, additional Bonferroni correction was used to adjust for multiple comparisons. Levene's test was used to check the equality of the variances of the different groups. In case of non-normality, data were log transformed. In case of non-normality of log transformation, data were evaluated using Kruskal-Wallis nonparametric test and Mann-Whitney U test. Data were analyzed using SPSS version 16.0.2. A 2-tailed p value of < .05 was considered to be statistically significant and 95% confidence interval (CI) was given.

The Effect of Sternocleidomastoid Muscle Cleavage
Twenty-seven patients underwent cleavage of the sternal attachment of the SCM and 56 patients did not. Only the Dutch population was selected to compare. Given that all included Dutch surgeons involved in this study were trained at The Netherlands Cancer Institute – Antoni van Leeuwenhoek Hospital, the surgical technique was assumed to be similar. In addition, intra-observer bias was known to be more than adequate in the Dutch population within all measurements (unpublished data). No subdivision was made in whether a neck dissection was carried out. Independent samples t-tests were performed to compare the peristomal dimensions in these 2 groups. Levene's test was used to check the equality of the variances of the different groups. When equality of the variances was not met, data were evaluated using a Mann-Whitney (Wilcoxon rank sum) nonparametric test.

The Effect of Age and Time after Operation
To determine whether there was a correlation between the age of the patient and peristomal geometry, and whether there was a correlation between the time after operation and the peristomal geometry, Pearson correlation coefficients were used. In case of non-normality, Spearman rank correlation coefficients were used. Additionally, visual scatterplots and boxplots were made in order to visualize a possible correlation.

The Effect on Pharyngeal Reconstruction
To investigate whether or not there was a difference in peristomal geometry when pharyngeal reconstruction was performed after total laryngectomy, the groups of patients who underwent a reconstruction (n = 21) and the group of patients who did not (n = 96) were compared with the independent samples t-test. Levene's test was used to check the equality of the variances of the different groups. If equality of the variances was not met, data were evaluated using the Mann-Whitney nonparametric test.

The Effect of Neck Dissection
To determine if there was a difference in peristomal anatomy in the group of patients who

had and who had not undergone neck dissection, 3 groups were distinguished: patients who did not have a neck dissection (n = 36), patients who had a 1-sided neck dissection (left, n = 9, in addition to right, n = 8), and patients who had a bilateral neck dissection (n = 67). These groups were compared using 1-way ANOVA. Furthermore, ANOVA was also used to compare the change in peristomal geometry as a result of the type of neck dissection; radical (n = 16), modified (n = 13), selective (n = 24), and absence of lymph node dissection (n = 24).
In case of significant results, additional Bonferroni correction was used to adjust for multiple comparisons.
Levene's test was used to check the equality of the variances of the different groups. If equality of the variances was not met, data were evaluated using a Kruskal-Wallis nonparametric test.

The Effect of Radiotherapy
To investigate whether or not there was a difference in peristomal geometry when radiotherapy was performed after total laryngectomy, the group of patients who had undergone radiotherapy (n = 70) and the group of patients who had not (n = 57) were compared using the independent samples t-test. Levene's test was used to check the equality of the variances of the different groups. If equality of the variances was not met, data were evaluated using Mann-Whitney nonparametric test.

Results

General
One hundred and ninety-one patients were included in this study, of whom 10.4% were women. The average age was 65.5 years (range: 35-92 years). In total, 49.6% used an adhesive stoma patch on a regular (daily) basis. In the group of regular daily adhesive stoma patch users, the mean duration *in situ* was 25.6 hours (range: 0-7 days). The overall anatomy of the trachea opening and peristomal soft tissue is described in Table 2, as stated by Van der Houwen *et al.*[43]

The Effect of Number of Removed Trachea Cartilage "Rings" during Laryngectomy
With respect to the geometry *H*, the overall significance was 0.004. Paired comparisons using Bonferroni corrected result demonstrated a significant difference between the group with 2 removed trachea rings and 3 trachea rings. The horizontal stoma diameter *H* was an average 1.87 mm smaller (p = .01; 95% CI, 0.40–3.33 mm) in the group with 3 removed trachea rings. Next, paired comparisons using Bonferroni corrected result demonstrated a significant difference between the group with 2 removed trachea rings and 4 trachea rings. The horizontal stoma diameter *H* was 3.33 mm in average smaller (p = .15; 95% CI, 0.51–6.16 mm) and the vertical stoma diameter *V* 4.16 mm smaller (p = .03; 95% CI, 0.25–8.08 mm) in the group with 4 removed trachea rings.

Table 2. Peristomal geometry, all measurements presented are in mm.

	Valid n	Mean	SD	95% CI
Circumference of the neck *Z*	166	367	33	310 - 420
Horizontal stoma diameter *H*	186	13	4	8 - 20
Vertical stoma diameter *V*	188	17	5	10 - 25
Distance fold-to-fold *F*	187	38	16	14 - 70
Distance stoma-to-chin *C*	186	35	15	10 - 114
Distance stoma-to-sternum *S*	188	24	9	10 - 40
Distance stoma to patients' right fold *R*	184	20	13	7 - 45
Distance stoma to patients' left fold *L*	185	19	9	8 - 40
Stoma depth hor. from upper side of stoma *u*	175	13	9	0 - 32
Stoma depth hor. from lower side of stoma *d*	175	12	8	0 - 27

Abbreviations: n, number of patients; CI, confidence interval; hor., horizontal

The Effect of Sternocleidomastoid Muscle Cleavage

The t-test did not show significant differences between the SCM-cleaved (n = 27) and non-cleaved group (n = 56). The Mann-Whitney test demonstrated a statistically significant difference in the distance from the stoma to the left fold *L* measurements (p = .02) and in the distance from the upper border of the stoma to the chin *Z* measurements (p = .00). No statistically significant difference was found between the cleaved and non-cleaved patients' group in the measurements *H, V, F, C, S, R, u,* and *d*.

The Effect of Age after Operation

A positive correlation was found between the age of the patient and the circumference of the neck (*Z*; i.e., there is an increase in this circumference with increasing age, as is shown in fig. 2). Pearson correlation coefficient was 0.19 (p = .02). Furthermore, age demonstrated a positive Pearson correlation between the depth of the upper border *u* of the stoma (CC = 0.20; p = .01) and the depth of the lower border *d* of the stoma (CC = 0.17; p = .04). No additional significant correlations were found.

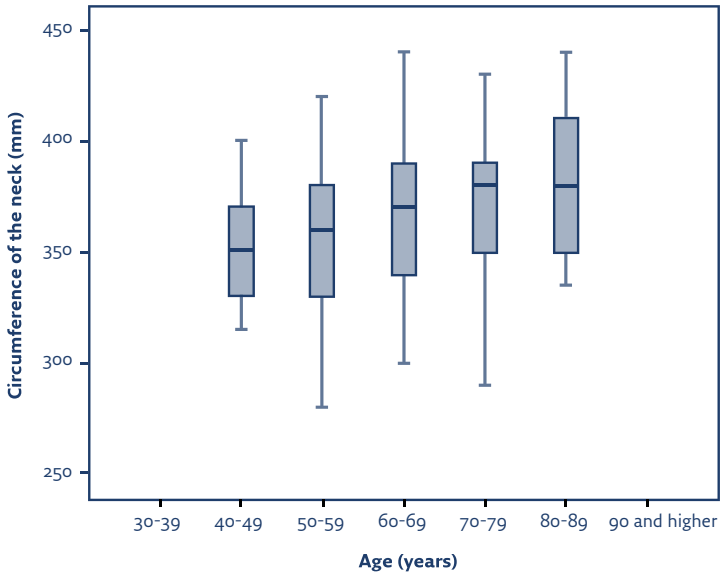

Figure 2. Boxplot demonstrating the relation between the age of the patient in years and the circumference of the neck. An increase in neck circumference is shown when the patient's age is higher.

The Effect of Time after Operation

A positive Pearson correlation between the time after operation and the distance from the upper border of the stoma to the chin *C* was demonstrated (CC = 0.182; p = .04). Furthermore, a negative Spearman's rho was shown between the time after operation and vertical diameter *V* of the stoma (CC = 0.18; p = .04). No significant correlation between the horizontal diameter of the stoma *H* and the time after operation was found.

The Effect on Pharyngeal Reconstruction

Statistically significant differences were found between the reconstructed and the non-reconstructed group. The group in which a pharyngeal reconstruction was carried out had a 5.70 mm larger depth of the upper side of the stoma (p = .01; 95% CI, 1.28 to 10.11 mm). No other influencing factors were demonstrated.

The Effect of Neck Dissection

A statistically significant difference was found between the non-dissected and the bilaterally dissected group. The non-dissected group showed a smaller horizontal stoma diameter (H) with a mean difference of 3.0 mm (p = .00; 95% CI, 1.2–4.7 mm) and a smaller vertical stoma diameter *V* with a mean difference of 3.7 mm (p = .00; 95% CI, 1.6–5.9 mm) as compared to the bilaterally dissected group. Furthermore, statistically significant difference was found between the unilaterally and the bilaterally dissected group. The bilaterally dissected group

showed a smaller vertical stoma diameter *V* with a mean difference of 3.64 mm (p = .02; 95% CI, 0.81–6.46 mm) as compared to the bilaterally dissected group. Finally, we compared the peristomal geometry as a result of the type of neck dissection (radical, modified, selective, and absence of lymph node dissection). Statistically significant differences were found between the non-dissected group and the selective lymph node dissected group: the selective lymph node dissected group showed a 3.49 mm larger horizontal *H* stoma diameter (p = .01; 95% CI, 0.61–6.78 mm) and a 4.01 mm larger vertical *V* stoma diameter (p = .01; 95% CI, 0.41–6.62 mm) as compared to the non-dissected group. Furthermore, the selective lymph node dissected group showed a 13.0 mm larger fold-to-fold *F* distance (p = .02; 95% CI, 1.15–24.86 mm) as compared to the non-dissected group. A statistically significant difference was found between the modified dissected group and the selective lymph node dissected group: the selective lymph node dissected group showed a 8.27 mm larger lower stoma *d* depth (p = .02; 95% CI, 0.73–15.82 mm) as compared to the modified dissected group.

The Effect of Radiotherapy

Statistically significant differences were found between the radiotherapy and the non-radiotherapy group. The group in which radiotherapy had been carried out had a 7.7 mm smaller distance between the SCM or skin folds *F* as compared to the non-radiotherapy group (p = .01; 95% CI, 1.5 to 13.8 mm). Furthermore, the group in which radiotherapy had been carried out had a 5.7 mm smaller distance between the right border of the stoma until SCM or skin fold *R* as compared to the non-radiotherapy group (p = .01; 95% CI, 2.1 to 9.7 mm). No other influencing factors were found.

Multivariate Analysis

Multivariate analysis was done in order to perform trade studies across all studied dimensions while taking into account the effects of all 6 variables. However, multivariate analysis was not able to signify these effects, due to missing variables.

Discussion

Although 3-dimensional evaluation of the (peri-)stoma using stereophotogrammetry[28] or laser scanning is well known, proven,[44] and was available in-house, we found this was of no practical and additional use in a worldwide study, as not all medical centers had access to these techniques and these techniques are just a storage medium between the anatomy and the actual measurement. Our proposed measurement protocol can also easily be carried out on a digital image of the patient. Our attempts to model the peristomal area were, therefore, performed in a relatively low-tech manner using a Vernier caliper and flexible ruler. This was sufficient to gain all 10 parameter values in a reliable way and omits 3-dimensional scanning which only results in an image but no parametrical values. It was assumed that when removing more trachea rings during total laryngectomy, more traction was carried out on the skin surrounding the stoma. However, in contrast to our expectations, the group of patients who had had 4 trachea rings removed did not have a deeper stoma (described in terms of *u* [depth

of the upper border of the stoma opening] and *d* [depth of lower border of the stoma opening]) when compared to the group with 2 trachea rings removed. This could suggest that the surgeon is not limited in a choice between 2 and 4 tracheal rings to be removed with respect to the stomal geometry. When evaluating the measurements of the SCM groups, it was assumed that incision of the sternal attachment of the SCM provided a flatter peristomal area, better suited to today's most used flat peristomal adhesive patches. Although the fold-to-fold distance seems larger, no significantly shallower stomas were found in the group where an incision of the SCM was made. As a result, it is not demonstrated that cleaving the SCM during laryngectomy will result in a geometrically flatter stoma.

Most clinicians familiar with laryngectomy patients know the phenomenon of a shrinking stoma after surgery. This occurrence could only be regained in our data in the vertical diameter *V* of the stoma. In addition, with respect to radiotherapy, this phenomenon could not be shown. However, it must be noted that measurements were carried out on a single outpatient day. Also, preceding dilatation of the stoma opening was not taken into account in this study.

Next, the influence of pharyngeal reconstruction on the peristomal geometry was studied. The lower border *d* did not significantly alter when comparing the 2 groups. However, the depth of the upper border of the stoma *u* proved to have significantly increased by as much as 5.70 mm. The measurement *u* was defined as the distance from the most superior edge of the stoma to the plane over the first interruptions lateral to the stoma. It must be noted that reconstruction is mainly performed above the level of the stoma. This could account for a larger upper stoma depth *u*. Furthermore, the location of the lower border of the stoma *d* is anchored by the framework of the sternum and clavicle and, therefore, seems not to move as much as the upper border.

A multivariate analysis was carried out for all factors, but due to missing variables, no additional results could be taken into account. Furthermore, in this worldwide multicenter study, no conclusions could be drawn upon stoma geometry between different ethnic groups.

Conclusion

These new data describing the tracheal opening and soft tissue contours around the trachea stoma do indicate the variety of influences affecting the peristomal area after total laryngectomy. Surprisingly, SCM cleavage did not seem to influence the depth of the peristomal area in this worldwide study. Therefore, it is not demonstrated that cleaving the SCM at the time of a laryngectomy will result in a geometrically flatter stoma. Furthermore, it seems that the number of removed tracheal rings during laryngectomy does not influence the depth of the stoma. Based on the results of our worldwide multicenter studies, a new adhesive automatic tracheostoma valve fixation patch is being developed. Our intent in developing a new biomedical device is not to adjust the patient to the device, but to engineer the device to fit the patient.

4

Investigating bi-stability

Functional Buckling Behavior of Silicone Rubber Shells for Biomedical Use. An example application.

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Submitted

Abstract

Background: The use of soft elastic biomaterials in medical devices enables substantial function integration. The consequent increased simplification in design can improve reliability at lower cost in comparison to traditional (hard) biomaterials. Functional bi-stable buckling is one of the many new mechanisms made possible by soft materials. The buckling behavior of shells, however, is typically described from a structural failure point of view: the collapse of arches or rupture of steam vessels, for example. There is little or no literature about the functional elastic buckling of small-sized silicone rubber shells, and it is unknown whether or not theory can predict their behavior. Is functional buckling possible within the scale, material and pressure normally associated with physiological applications? An automatic speech valve is used as an example application.

Method of Approach: Silicone rubber spherical shells (diameter 30 mm) with hinged and double-hinged boundaries were subjected to air pressure loading. Twelve geometrical configurations were tested for buckling and reverse buckling pressures. Data were compared with theory.

Results: Buckling pressure increases linearly with shell thickness and shell height. Reverse buckling shows these same relations, with pressures always below normal buckling pressure. Secondary hinges change normal/reverse buckling pressure ratios and promote symmetrical buckling. All tested configurations buckled within or closely around physiological pressures.

Conclusions: Functional bi-stable buckling of silicone rubber shells is possible with adjustable properties in the physiological pressure range. Results can be predicted using the proposed relations and equations.

Introduction

Elastic materials enable possibilities in mechanical design that cannot possibly be achieved with traditional hard materials. Soft materials used in biomedical applications are also mechanically more compatible with soft biological tissue than the traditional hard plastics and metals used for implants. For that reason, they are accepted very well by the body. Examples of elastomers that are used for implantable biomedical applications are silicone rubber, polyurethane (PUR) and a number of thermoplastic elastomers (TPEs) such as SIBS30[23] or even biological rubber-like materials.[24]

The mechanical properties of rubber create new applications. One such geometry that allows for new applications is the small spherical shell in its function as a bi-stable snap-through buckling shell. Due to their large elastic range they do not deform plastically during the snap-through process and are therefore able to buckle back (so called "reverse buckling"). This geometry can function as or within check valves, servo valves, bi-state mechanical memory cells and the like, as a separate part or within a rubber mechanism.

This paper describes how to predict the properties of such small bi-stable shells in theory and, with an example application, in practice.

Bi-stable buckling

When a spherical shell or cap as depicted in figure 1 is submitted to mounting external pressure, it may retain its spherical shape and only experience uniform compression.[45] When the pressure increases beyond a certain value, the spherical shape of the compressed shell may become unstable and buckling occurs. Snap-through buckling (snapping, buckling, oil-canning) is characterized by a visible and sudden jump from one equilibrium configuration to another equilibrium configuration (fig. 1, right). The load or pressure under which the shell buckles is called the "buckling load", whereas the "critical load" is designated as the load at which buckling occurs according to calculations from theoretical models[46,47] A system is considered bi-stable if it can exist in either of two stable states between which buckling can take place.

The stability of spherical shells has been studied for several decades.

Applications are found in structures such as steam pressure vessels, storage tanks[48] and church domes, for example, and typically cover structural failure: the collapse of arches or the rupture of pressurized vessels, etc.

Studies of metal spherical structures indicate that the stability of a spherical shell can be predicted by its geometry.[49-61] In literature, a distinction is made between "thick" and "thin" shells. Whereas thin shells are said to follow the predictions of "classical theory", calculating the critical pressures for thick shells "involves solving a fourth-order system of highly non-homogeneous differential equation".[62]

Recently the theory of bi-stable buckling has been used in the analyses of very small micro-lenses[63] or in nano-scale mechanisms.[64]

The mechanisms of highly elastic materials (materials with a low Young's modulus), such as silicone rubber, may be bi-stable and allow reversible buckling that can be repeated many times. Unfortunately, the structures investigated in literature are of a completely different

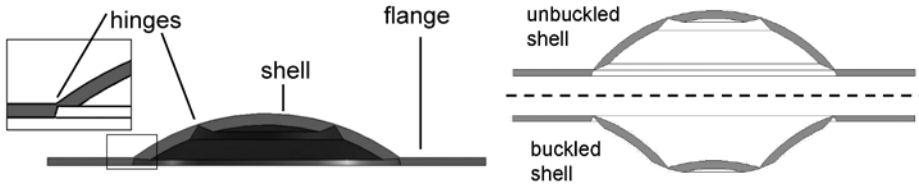

Figure 1. Cross-section of a specimen with secondary hinge (left) and a shell in normal condition (top right) and buckled (bottom right).

scale (decimeters to meters) and different materials (steel, concrete), compared to the biomedical applications of elastomer spherical shells, like the small silicone rubber shells used in a handsfree speech valve (see next section). Moreover, reverse buckling is currently not being studied at all, the properties of those materials rendering this impossible.
To find the optimal geometric configuration of a bi-stable silicone rubber shell, its buckling behavior must be known.

An example application: A handsfree speech valve

An application of such a bi-stable valve is the iValve,[65] a "handsfree speech valve" for patients whose larynx has been excised as a remedy against cancer of the throat (laryngectomy patients). These patients, missing their speech organs, now breathe through an artificial opening in the front of the neck (a tracheostoma) and are routinely surgically fitted with a check valve, connecting the trachea to the esophagus: at occlusion of the stoma with a finger, exhaled air is now pressed through that check-valve, provoking vibration of the soft esophageal tissue, and can then be articulated into speech by the user (fig. 2).
To avoid the use of a finger, handsfree speech that can be put into a "speech mode" by a short spurt of inhalation is currently under development.
The iValve in this example is made entirely of silicone rubber. This allows the production of complex parts that integrate many functions, so the device consists of only two assembly parts. One part features a bi-stable shell that, by virtue of its inward or outward buckled shape, blocks or releases a central duct (fig. 3). For working details see Chapter 5.[65] The bi-stable shell used in this design was derived from the data presented in this paper.

Research question

This study was set up to investigate the bi-stable buckling behavior of silicone rubber shells and to see if the properties fall within the range useful for application in medical devices.
For a shell of given material properties and size, we investigated how shell thickness, shell height and the position of a second hinge influence the buckling pressure and reverse buckling pressure. Reverse buckling was thought to be below normal buckling due to the buckling pre-tensions stored within the material. A secondary hinge was thought to promote symmetrical buckling (along the axis of rotational symmetry) and change the normal/reverse buckling pressure ratio. Next, we investigated if buckling pressures can be predicted by a simple method.

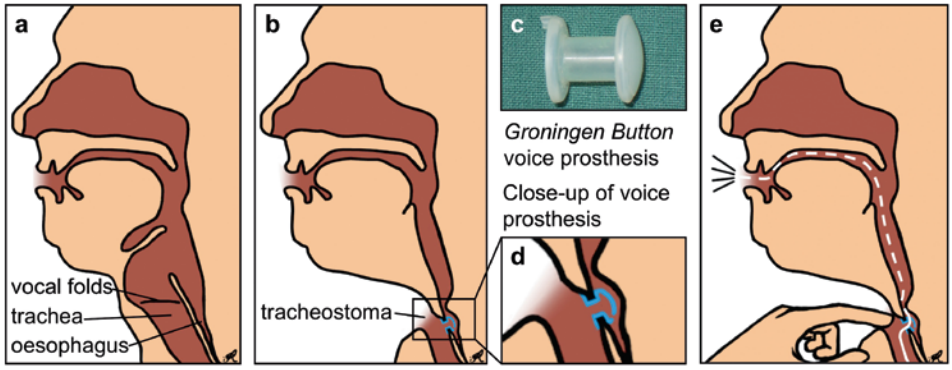

Figure 2. Laryngectomy: A healthy patient (a) and a laryngectomy patient (b) missing the vocal folds. A shunt valve (c) is routinely implanted between the oesophagus and trachea (d) so that after inhalation and closure of the stoma (e), all exhaled air is rerouted via the oesophagus, which will vibrate: an artificial voice.

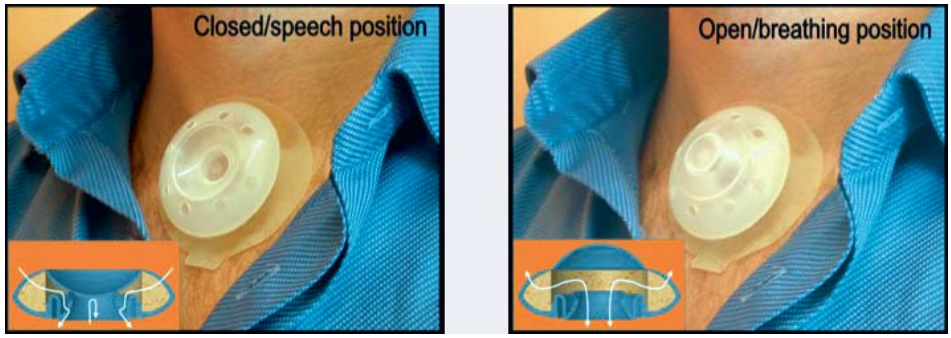

Figure 3. Automatic speech valve featuring a bi-stable buckling (double hinged) shell as its switching mechanism.

Materials

Twelve different silicone rubber spherical shell configurations with a diameter of 30 mm (figs. 1 and 4) were molded from silicone rubber (Z001, Principality Medical Limited, Rogerstone, UK) with varying shell height *H*, shell thickness *h*, and, if a secondary hinge was present, varying secondary hinge angle φ (fig4. and table 1). According to the definitions used in classical theory, shells are considered deep when $H/b \geq 0.167$ and thick when $h/a \geq 0.05$.[61] Four shells could be considered thick and eight thin; all were deep shells.

From every configuration, 6 identical specimens were tested. Most literature uses the shells radius a as a parameter. But since shell height *H* is more easily measured within a physical shell, we used *H* throughout this study and derived the parameter a when needed. All our shells had a fixed diameter *b* of 30 mm, thus a and *H* were directly mathematically related.

Notations
a = shell radius
b = base radius
H = shell height
h = shell thickness
θ = semi-angle
(φ = secondary hinge angle)
E = Young's modulus
P_{static} = average static buckling pressure
$P_{dynamic}$ = average dynamic buckling pressure
$P_{static,\ reverse}$ = average static reverse buckling pressure
$P_{dynamic,\ reverse}$ = average dynamic reverse buckling pressure

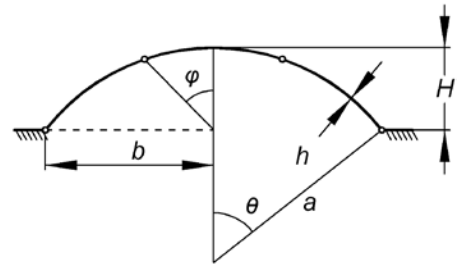

Figure 4. Notations and Cross-section of a spherical shell with a secondary hinge.

Reproducibility
The reproducibility of the shells was determined by measuring the shell thickness *h*, of each specimen on different locations with a micrometer. The shell height *H*, was verified using a Vernier height gauge.

E-Modulus
The tangent modulus in tension of the used material was not specified; therefore, we determined it around the expected strain. This strain was calculated as 25% for the shallowest and 52% for our deepest shell, when buckled. Ten standard tensile specimens (ISO 37 Dumb-bell type 2) were molded from the used silicone rubber.[66] The specimens were tested in a tensile test machine (Zwick/Roell TC-FR2, 5TS.D09, 2.5kN test machine. Force accuracy 0.2%, positioning accuracy 0.01 mm; Zwick/Roell Nederland, Venlo, The Netherlands) with a speed of 1 mm/min. The tangent modulus for strains between 10 – 60 % turned out to be constant at 1.85 MPa (± 0.04 MPa). A Poisson Ratio ν = 0.48 typical to this type of silicone rubber was assumed.

Methods

Figure 5 depicts the experimental set-up. All specimens were fixated with their dome downward by clamping the specimen flange between the supporting plate and the retaining ring. The circumferential hinge was free to move (± 0.5 mm free space). A computer-controlled valve (pressure regulator QB2TFEE002 with volume booster R0002-B, Proportion-Air, McCordsville, USA) regulated the pressure within the volume enclosed by the specimen and the pressure system. The pressure under the specimen was monitored continuously using a calibrated pressure sensor (DSY002, Proportion-Air, McCordsville, USA), and plotted against time. Three measurements were performed per specimen and the results were averaged.

Table 1. Dimensions of the different configurations (S single and D double hinge).

Configuration (n)	Shell height H (mm)	Shell thickness h (mm)	Secondary hinge angle φ (degrees)	Shell radius a (mm)	Base radius b (mm)	Semi-angle φ (degrees)	Shell thickness h/a ratio	Shell depth H/b ratio
S1 (6)	3.0	1.0	--	39	15	22.6	0.026	0.200
S2 (6)	5.0	1.0	--	25	15	36.9	0.040	0.333
S3 (6)	5.0	0.7	--	25	15	36.9	0.028	0.333
S4 (6)	5.0	1.3	--	25	15	36.9	0.052	0.333
S5 (6)	7.0	1.0	--	20	15	50.0	0.050	0.467
D1 (6)	3.0	1.0	65	39	15	22.6	0.026	0.200
D2 (6)	5.0	1.0	35	25	15	36.9	0.040	0.333
D3 (6)	5.0	0.7	65	25	15	36.9	0.028	0.333
D4 (6)	5.0	1.0	65	25	15	36.9	0.040	0.333
D5 (6)	5.0	1.3	65	25	15	36.9	0.052	0.333
D6 (6)	5.0	1.0	75	25	15	36.9	0.040	0.333
D7 (6)	7.0	1.0	65	20	15	50.0	0.050	0.467

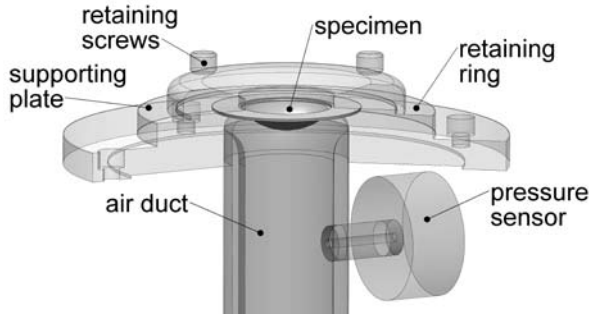

Figure 5. Cut-through of the experimental respiratory set-up. The spherical specimen is clamped under the retaining ring, dome downward.

Normal and reverse buckling under static loading

After placing the specimen, static as well as dynamic loads were applied by increasing the pressure in the air feed below the spherical shell. Static buckling pressures were determined by increasing the pressure slowly and gradually (step size 0.025 kPa and step duration 2.0 seconds) until snap-through buckling occurred. The static buckling pressure was defined as the maximum pressure under the spherical specimen just before buckling (characterized by a sudden drop of pressure). Static buckling pressures were also obtained for reverse buckling (i.e., buckling back to the original state), by applying pressure load on the convex surface of the specimens in their inverted (buckled) state.

Normal and reverse buckling under dynamic loading

Dynamic buckling pressures were determined by applying fast step pressure loads (rise time < 0.15 seconds). The dynamic buckling pressure was defined as the step pressure at which buckling occurs instantly after application of the pressure load, as used by previous experiments and analyses.[49,51,54,57,58,61,67,68] Dynamic buckling pressures were also determined for reverse buckling.

Statistical methods

The influence of the loading condition and the geometrical variable on the buckling pressure was analyzed using SPSS version 16.0 (SPSS Inc. Chicago, Illinois, USA). Differences in buckling pressure between static and dynamic loading were investigated by multilevel linear model analysis with a significance level of 0.05. To establish the relation between the variables and the buckling pressure, curve estimations were done using curve fitting. Linear relations were checked for fit by their R^2 value in Microsoft Excel 2003 (Microsoft Inc., Pasadena, USA).

Theory

Classical theory as found in literature gives a simple equation for the static critical buckling pressure, which is said to be only valid for thin, shallow shells.[47]

$$P_{cr} = \frac{2E}{\sqrt{3(1-\nu^2)}} \left(\frac{h}{a}\right)^2$$ (see fig. 4 for all parameters)

Equation 1

This equation was later adjusted to fit experimental results.[56] The resulting equation was found to be a good predictor when limited to $0.0005 \leq h/a \leq 0.0025$ and $20° \leq \theta \leq 60°$.[27]

$$P_{cr} = (1 - 0.008758(\theta - 20deg))\left(1 - 0.000175\ \frac{a}{h}\right)(0.3E)\left(\frac{h}{a}\right)^2$$

Equation 2

Our shells under investigation had semi-angles $22.6° \leq \theta \leq 50°$, which is inside the suggested range, but thickness ratios $0.025 \leq h/a \leq 0.052$, which are outside the suggested range. However, Litle (1964) concludes that buckling behavior of shells is not affected by a 100% change in geometric scale and, by using dimensional analysis, shows that buckling pressure is independent of geometric scaling.[69]

Results

The shell thickness showed deviations of up to 6% from the mean in the thinnest shells (D3 and S3) and up to 2% in the thicker ones (D4 and S5).
The shell height deviated up to 5% from the mean in the shallower specimens (D1 and S1), and 11% in the deeper ones (D5 and S7).
The static and dynamic buckling and reverse buckling pressures (and their ratios) are presented in Table 2.
Multilevel analysis indicated that there was no significant difference in buckling pressure between static and dynamic loadings ($P = 0.075$). Therefore, we only present results for static buckling from this point onward. Buckling pressure values for reverse buckling are always lower than those for normal buckling (see table 2, last column, $P_{static,\ reverse} / P_{static}$).
Subsequently we investigated the relation between buckling pressure, shell height *H*, shell thickness *h*, and the presence of the secondary hinge and its angle.
Figure 6, first graph, suggests that the relation between buckling pressure and shell height *H* is linear. Best fit through observed means gives a linear relation: $P = 0.6867H - 1.597$, ($R^2 = 0.9918$) for normal and $P = 0.5454H - 1.3899$, ($R^2 = 0.9813$) for reversed buckling shells with a secondary hinge at 65°. Figure 6, second graph, suggests a linear relationship:

Table 2: The buckling and reverse buckling pressures of single hinge (S) and double hinged (D) shells. All pressures in kPa. The setup did not allow for testing dynamic reverse buckling below 0.5 kPa; the table states "too low".

Configuration (n)	P_{static} (SD)	$P_{dynamic}$ (SD)	$P_{static,\ reverse}$ (SD)	$P_{dynamic,\ reverse}$ (SD)	$P_{static,\ reverse} / P_{static}$
S1 (6)	0.53 (0.08)	0.51 (0.05)	0.18 (0.06)	too low	0.34
S2 (6)	1.63 (0.24)	1.68 (0.26)	0.76 (0.11)	0.86 (0.08)	0.47
S3 (6)	0.97 (014)	0.96 (0.12)	0.38 (0.05)	too low	0.42
S4 (6)	2.52 (0.08)	2.57 (0.07)	1.11 (0.06)	1.15 (0.05)	0.44
S5 (6)	3.91 (0.12)	4.03 (0.21)	1.89 (0.06)	2.10 (0.05)	0.48
D1 (6)	0.54 (0.12)	too low	0.33 (0.05)	too low	0.62
D2 (6)	2.05 (0.15)	2.06 (0.17)	0.83 (0.07)	0.86 (0.06)	0.40
D3 (6)	0.85 (0.19)	0.88 (0.18)	0.52 (0.10)	too low	0.61
D4 (6)	1.69 (0.15)	1.70 (0.17)	1.16 (0.10)	1.23 (0.10)	0.69
D5 (6)	2.03 (0.17)	2.04 (0.13)	1.38 (0.14)	1.48 (0.17)	0.68
D6 (6)	0.98 (0.07)	0.95 (0.10)	0.60 (0.04)	too low	0.61
D7 (6)	3.28 (0.33)	3.48 (0.33)	2.52 (0.25)	2.64 (0.26)	0.77

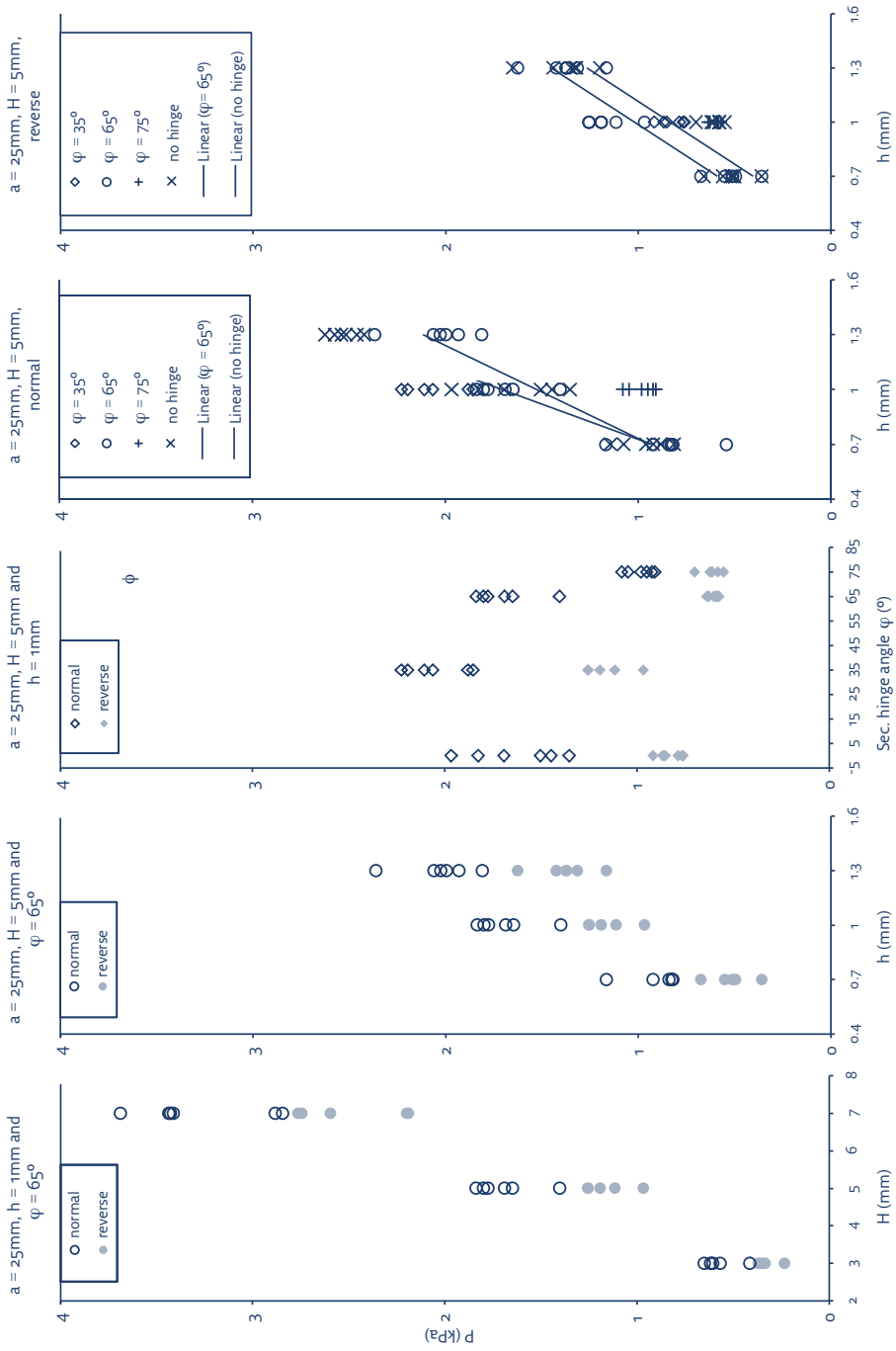

Figure 6. Buckling and reverse buckling pressures for different shell height H, shell thickness h and secondary hinge angle φ with parameters kept constant mentioned above the graph. The outmost right two graphs show the influence of shell thickness h in the buckling pressure for different hinge angles φ and for the shell without a secondary hinge (presented as o° in the graph), with linear trend lines through configurations of the same hinge angle.

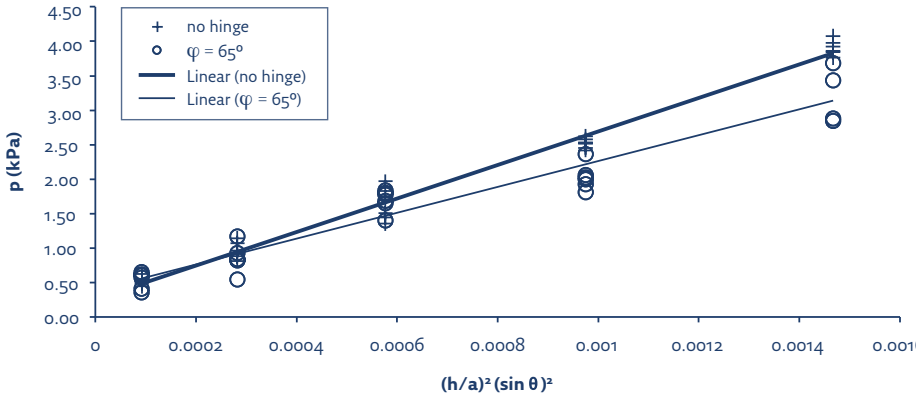

Figure 7. Relation between buckling pressure P and (h / a)2 * (SIN θ)2 for normal buckling with shells without a secondary hinge and with a φ = 65°.

P = 1.963h – 0.4374, (R^2= 0.9932) for normal buckling, and P = 1.4389h - 0.4181, (R^2 = 0.9244) for reverse buckling shells with a secondary hinge at 65°. Figure 6, third graph, shows that the buckling pressure in relation to the secondary hinge angle φ is non-linear. It can be debated whether no hinge can be interpreted as a hinge angle of either 0° or 90°; in neither case do the data show a consistent relation with the other angles. Generally, the buckling pressure decreases with larger secondary hinge angles. Although the difference between the normal buckling pressure at 65° and 75° is significant, there is no significant difference with inverse buckling. It was also visually observed that, by and large, buckling occurs more symmetrically when a secondary hinge is used than in the case of similar geometries without this secondary hinge.

Discussion

Effect of geometry on the buckling pressure

Taeprasartsit and Tao[61] show that the buckling pressure is a function of h/a and θ alone and from the classical formula (equation 1) we know that the static buckling pressure is proportional to $(h/a)^2$. From these relations and a suggestion by an anonymous reviewer, we found a linear relation between P and $(h / a)^2$ * (SIN θ)^2 = $h^2 b^2 / a^4$ (R^2 = 0.9864 for no hinge and R^2 = 0.9278 for φ = 65° (fig. 7)) that even encompasses the thick shells, not covered by the classical theory.Taeprasartsit and Tao also state that neither the Young's modulus E nor density ρ have an effect on P_{cr} although E is in the classical equation (equation 1). As we only used one type of material, this relation was not investigated in our study.

Substitution with this relation and the trend line from figure 7 results in our own buckling equation for shells without secondary hinges:

Table 3. Summed squared difference (SSD) between observed values and predicted values from different theories. Lower SSDs mean higher conformity to observation.

Configuration (n)	Observed Mean (SD)	Classical theory (kPa)	Klöppel & Jungbluth (kPa)	New equation (kPa)
S1 (6)	0.53 (0.08)	6.94	0.20	0.03
S2 (6)	1.63 (0.24)	31.05	4.91	0.29
S3 (6)	0.97 (0.14)	5.42	2.21	0.08
S4 (6)	2.52 (0.08)	99.12	9.35	0.10
S5 (6)	3.91 (0.12)	28.57	50.12	0.10
SSD of all observations		171.10	66.79	0.59

$$P_{cr} = 2433.3\left(\frac{h}{a}\right)^2 (\sin\theta)^2 + 0.2589$$

Equation 3

and an equation for shells with a secondary hinge at $\varphi = 65°$:

$$P_{cr} = 1877.1\left(\frac{h}{a}\right)^2 (\sin\theta)^2 + 0.386$$

Equation 4

Table 3 compares Equation 3 to the equations from classical theory and Klöppel & Jungbluth[56] and shows the summed squared differences (SSD) between predicted values and all observations. Lower SSDs mean higher conformity between measured and predicted data. Table 3 shows that our new equation has by far the lowest SSD of 0.59 and thus the best predictive value.

Static/dynamic buckling

Multilevel linear model analysis shows that there is no significant difference between static and dynamic buckling (P = 0.093). This does not concur with previous observations[54,57] that dynamic buckling loads of structures are generally lower than the corresponding static buckling load. Possibly our generated dynamic load is not fast enough.

Normal/reverse buckling

Reverse buckling pressures are always below normal buckling pressures for the same shell. Since the $P_{reverse} / P_{normal}$ ratio is dependent on the geometry of the shell (table 2), the secondary hinge angle, for instance, might be used to change this ratio for specific applications. The operation of the automatic speech valve in the example application is based upon different buckling pressures for its open and closed states.

Secondary hinge

Secondary hinges promote symmetrical buckling as compared to shells without a secondary hinge. For applications (as in our example application) where the shell is to "land" on a designated structure after buckling, symmetrical buckling could be (and in our example proves to be) essential for proper closure.

Depending on the application and design, the shell surface encompassed by the secondary hinge could be substituted in the case of a plain surface, thereby decreasing the effective height of the shell. Furthermore, since buckling pressure dependence on secondary hinge angles is different in the case of normal end reverse buckling, this parameter could be used to alter the normal/reverse buckling pressure ratio.

Biomedical applications

Physiological ranges of air pressures in the human respiratory system vary between 0 and 5 kPa;[70] human blood pressures range from approx. 7 to 20 kPa.[71]

The observed range of buckling pressures is 0.5 to 4.0 kPa, which is in the order of magnitude of these physiological pressures.

Therefore the biocompatible silicone rubber we used is an excellent material for the manufacture of bi-stable buckling shells, as in an automatic speech valve, for example. A study by El Fray et al. shows that for prolonged implantation, materials like SIBS30 are even more resilient to fatigue and are available in the same material property range as silicone rubber.[23]

Conclusions

Both shell thickness and shell height strongly influence the buckling pressures of the shells. Increasing the shell thickness or shell height seems to increase the buckling pressure linearly. Buckling theory seems to overestimate buckling pressures. A secondary hinge promotes symmetrical buckling in the shell design, lowers the effective height of the shell, and changes normal/reverse buckling pressure ratios, all of which can be useful in certain applications. Reverse buckling pressures are always lower than normal buckling pressures. All investigated shells buckle and reverse buckle within or closely around our physiological ranges.

The buckling pressure turns out to be linearly proportional to $(h / a)^2 * (\mathrm{SIN}\ \theta)^2$ which, in combination with a linear fit, results in equations that allow us to calculate buckling pressures to geometrical parameters. Since buckling pressures are scale independent,[69] we now have all the tools needed to design rubber shells suited to any application, using only limited experiments.

The functional bending, buckling and deformation of soft, elastic materials will enable new possibilities in medical applications with substantial function integration, and increased bio-compatibility.

5

In vitro evaluation of the iValve v1

In Vitro Evaluation of the iValve: A Novel Handsfree Speech Valve

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Abstract

Objectives: We performed an in vitro evaluation of a novel, disposable, automatic handsfree tracheostoma speech valve for laryngectomy patients based upon the principle of inhalation. The commercially available automatic speech valves close upon strong exhalation and open again when the pressure drops. This method makes long sentences or pauses difficult. The novel iValve is designed to allow almost natural speech, with mid-sentence pausing and whispering.

Methods: The inhalation closing flows and exhalation opening pressures of 6 iValve prototype versions at different settings were compared with physiological values. The airflow resistance at inhalation was compared to physiological and commercial valve values.

Results: The iValve prototypes showed a flow and pressure range in concordance with the physiological values in the literature. The airflow resistance in the breathing mode was within the physiological airflow resistance range, yet above the values from the two commercial valves. The resistance in speaking mode was above the physiological airflow resistance range.

Conclusions: In vitro tests show that the iValve versions can be selected and adjusted to operate within physiological range. The airflow resistance in the breathing mode is good. In speaking mode inhalation should, and can, be decreased. The iValve should offer the patient a more intuitively useable alternative with more dynamic speech. Its low cost allows disposability and wider use.

Introduction

Tracheostoma valves for speech restoration are known to improve the overall quality of life in laryngectomy patients.[9,10] In particular, handsfree speech using automatic tracheostoma speech valves (ASVs) is very well received by patients.[9,10] These valves have been on the market for some time.[6,11,12] The most commonly used ASVs are the Atos Provox FreeHands adjustable tracheostoma valve (Atos Medical AG, Hörby, Sweden), the Blom-Singer ATSV standard and ATSV II adjustable tracheostoma valves (Helix Medical, Carpinteria, USA) and the Adeva Window (Adeva Medical, Lübeck, Germany). All commercially available ASVs close upon strong exhalation and open again when the pressure drops. This method wastes valuable air during closure, leaving less air for longer sentences without using additional breaths. During speech, pressure should be maintained to prevent the valve from opening again, making pausing between words impossible. The resulting sentences are often short, with frequent breaths of air in between, leaving patients feeling out of breath and quickly fatigued.

A new ASV principle based upon inhalation[13–15] has already been demonstrated. This principle eliminated the problems associated with ASVs based upon the principle of closure upon exhalation. Major improvements of this mechanism, compared to exhalation ASVs, are the possibility of prolonged speech, the possibility of coughing without removing the valve, and the prevention of wasting exhaled air in closing the valve.[14] The newly developed iValve (figs. 1C and 2) introduced in this paper is based upon this principle. The current 3-part design is targeted to decrease manufacturing costs in comparison to existing valves, possibly allowing use as a one-use-only disposable device.

This study investigates the feasibility of the iValve for clinical use by comparing its operational characteristics to physiological values. First, we checked if the closing flows and opening pressures of the iValve could be independently controlled within the physiological pressure and flow ranges that can be produced by laryngectomy patients. Second, we investigated whether the airflow resistance coefficient (ARC) at inhalation in both speaking and breathing modes is comparable to the healthy physiological situation and to that with two commercially available, and widely used, valves.

Materials and methods

iValve

The iValve is composed of only 3 parts that are assembled to form a single unit. The device comprises a cap, a cap seat, and a foam-like heat and moisture exchange (HME) filter in between (fig. 2).

In order to start speech, the patient closes the iValve by strong inhalation (just exceeding normal breathing flows). In this speech position, the patient can speak (fig. 2B) almost naturally, pausing and whispering at will. The cap is the actual switching valve. The cap can buckle from the outward, breathing mode (fig. 2A), to the inward (fig. 2B), speaking mode, in which it effectively closes off the central breathing conduit, rerouting all exhaled air for speech. The valve will remain in the speech position, and inhalation is only possible via a different route

A B C

Figure 1. Atos-Provox FreeHands HME automatic exhalation valve (with heat and moisture exchange (HME) filter) A) on patient and B) on patch next to C) iValve automatic inhalation valve (without HME).

through the inside inhalation holes (fig. 2C, green arrow), around the central breathing conduit of the cap. To be closed at exhalation, the inside holes are covered with a circumferential flap on the patient side of the cap seat, which closes off the holes at exhalation when the circumferential flap is pressed against the inhalation holes by air pressure during speech.
The patient can switch the valve back into the breathing mode by strong exhalation, exceeding the phonation pressure (fig. 2A): the cap buckles outward, and air can freely flow through the central breathing conduit, the HME, and the breathing holes. All air – during speech only at inhalation, and during breathing at both inhalation and exhalation – is drawn through the HME filter. Exhalation for breathing is possible in the speaking mode, but will not be "voiceless"; for silent and higher-flow exhalation, the patient will have to exhale strongly and thus switch the valve back to the breathing mode.
The opening pressure and closing flow can be adjusted to suit the individual patient. The number of air holes is designed to determine the closing flow of the iValve. The pressure required to open the iValve is designed to be varied by changing the cap thickness. At the time of this study, an HME-filter that was also suitable for the in vivo follow-up study was not yet available. Therefore, all opening and closing tests were performed without this filter. The absence of the HME filter will not influence opening pressures, since the iValve is then closed and there is no airflow at all. It is however, likely to influence the closing flow. The HME was not considered essential to this stage of evaluation of the novel iValve prototype design.
For the opening pressure and closing flow measurements, 18 iValve caps with 6 different cap thicknesses (1.10, 1.17, 1.24, 1.36, 1.40 and 1.44 mm; 3 identical specimens per thickness) were molded from silicone rubber (Zoo1, Principality Medical Limited, Newport, UK). All caps were tested in conjunction with a single cap seat, molded from the same material.
For comparison of airflow resistance at inhalation, one Atos Cassette manual speech valve and one Atos FreeHands automatic speech valve (with the blue valve flap) served as references.

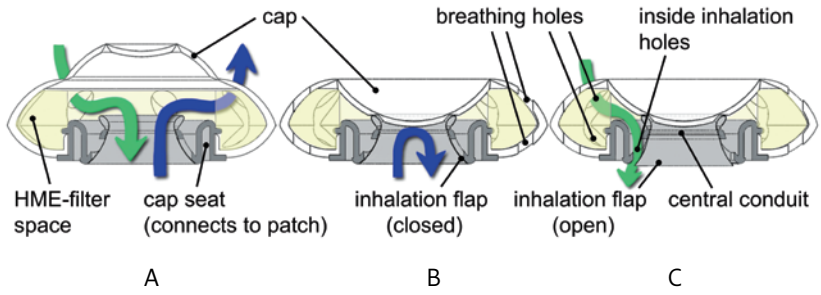

Figure 2. iValve handsfree inhalation valve prototype (with space for the HME-filter in yellow) A) in open, breathing position, B) in closed, speaking position, and C) at inhalation (inhalation flap open) in closed, speaking position. Inhaled air indicated by green, exhaled air indicated by blue arrows.

Test set-up

The valve's response characteristics were tested on a pneumatics simulator, a device that simulates the physiological pneumatics of respiration, speech and coughing (fig. 3). The device comprised a testing table with a standard Atos Provox connector to which all valves were attached for testing.

A computer-controlled valve (pressure regulator QB2TFEE002 with volume booster R0002-B, Proportion-Air, McCordsville, USA) controlled the pressure in the testing chamber under the testing table with the valve. To test the valves under inhalation, the testing table with the valve attached was mounted upside down on the pressure chamber. The pressure in the testing chamber under the valve was monitored continuously with a pressure sensor (DSY002, Proportion-Air), and plotted against time. The flow through the testing table was monitored continuously by means of a Lilly-type wire gauze flow head (AD Instruments MLT300L, Oxford, UK) that was connected to a differential pressure transducer (Honeywell DC001NDC4, Freeport, USA) and plotted against time as well.

Investigation of closing flow and opening pressure

All iValves were submitted to a dynamic pressure pattern (rise time less than 0.15 seconds). For this, a block-shaped pressure pattern (1.5 second plateau) was generated, each time with increased pressure (0.1-kPa steps starting at 1 kPa) until the flow or pressure at which the valve switched was established.

Every cap (6 thicknesses x 3 identical specimens) was tested with 4, 6, 8 and 10 air holes, punched in the edge of the cap, and tested 3 times at both inhalation (determining closing flow) and exhalation (determining opening pressure). Thus, 216 closing pressure values and 216 opening flow values were collected in total.

Investigation and comparison of air resistance

The ARC used throughout this paper is defined as the pressure difference divided by the squared flow as suggested by Verkerke et al. This definition of ARC yields a coefficient independent of airflow. [72]

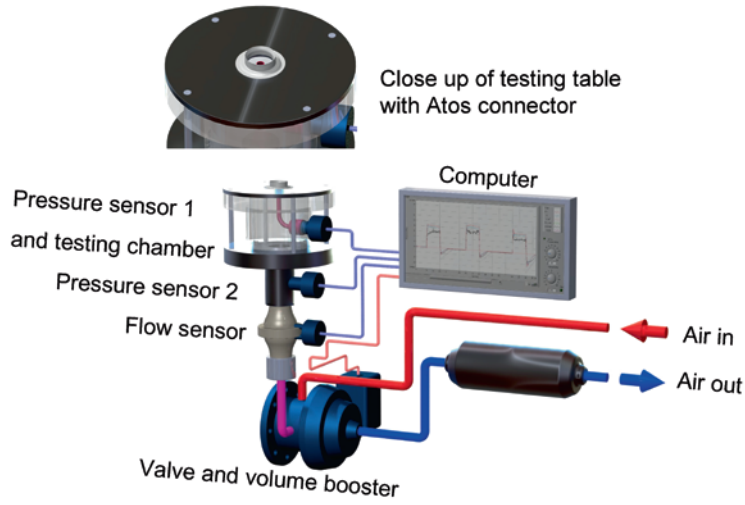

Figure 3. Pneumatics Simulator and close-up of Atos patch on top the testing table (top).

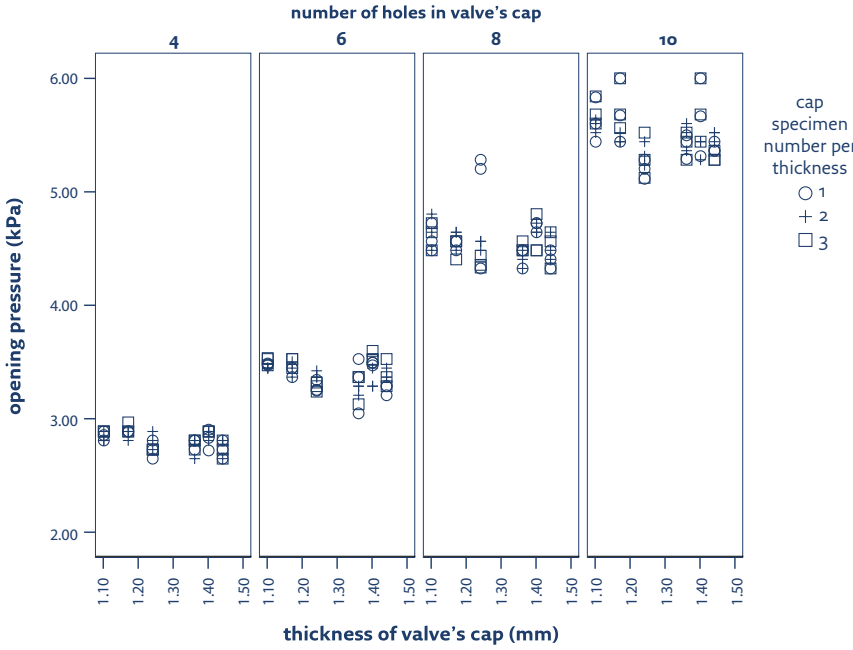

Figure 4. Airflow (y-axis) at closing in 18 iValve specimens as function of cap thickness (x-axis) and number of air holes in the cap (consecutive panels). Three identical iValve cap specimens per thickness were each tested 3 consecutive times.

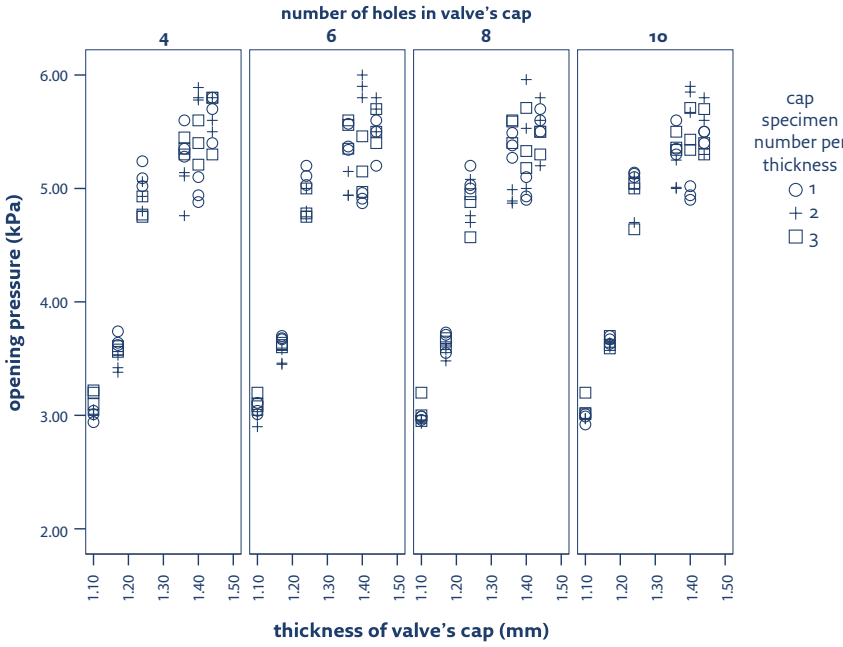

Figure 5. Air pressure at opening (y-axis) in 18 iValve specimens as a function of cap thickness (x-axis) and number of air holes in the cap (consecutive panels). Three identical iValve cap specimens per thickness were each tested 3 consecutive times.

For this experiment, 6 iValves with only 1 cap thickness (1.24 mm, representing the average thickness) were used. The iValve's ARC was measured in both breathing (open) and speaking (closed) modes.

The ARCs at inhalation of the iValve, Atos Cassette and Atos FreeHands were tested using the same pneumatics simulator setup. The airflow through the valves was slowly increased from 0.6 to 3.0 L/s, and the air pressure within the valve was registered at 0.3-L/s intervals. The ARC was calculated from these data as the pressure difference (between the atmospheric and the testing chamber pressure) divided by the squared flow through the valve. For fair comparison, the iValve was equipped with an HME filter in the designated cap cavity, cut from Atos HME filter foam. Although this filter made it possible to test the ARC, it was not yet suitable for safe *in vivo* testing.

Statistics

The relation between flows, pressures, cap thickness, and number of holes was investigated using a linear mixed-effects model generated with a restricted maximum likelihood approach. In all tests, an alpha level of 0.05 was used to indicate significance.

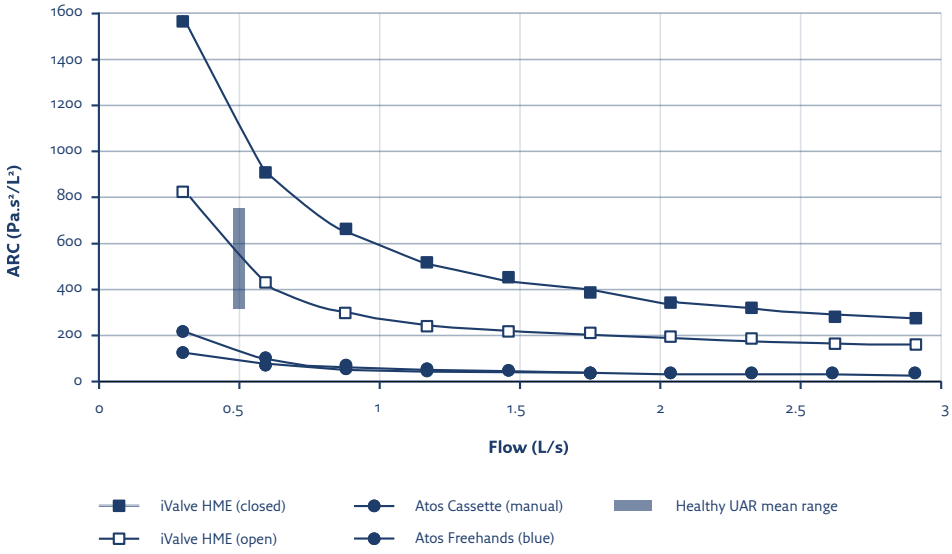

Figure 6. Airflow resistance coefficient (ARC; Pa · s2/L2) at inhalation versus flow (in L/s) for iValve (both open and closed), Atos Freehands, and Atos Cassette. Natural upper airway resistance mean range is 0.5 L/s. [73]

Results

Closing flow and opening pressure of iValve

Figure 4 shows the closing flow (from breathing to speaking mode) as a function of the thickness of the valve's cap (within a single panel) and as a function of the number of holes (between panels). When air holes are added, a clear increase in airflow can be observed, whereas the airflow seems unaffected by the valve cap thickness. The statistical mixed effect analysis confirmed that the number of holes has a significant ($\alpha = 0.05$) linear influence on the closing flow ($p < 0.001$), whereas the closing flow is not significantly influenced by cap thickness ($p = 0.056$). Any 2 added holes increased the closing flow by 1.16 L/s. The closing flow range is 1.8 to 6.0 L/s.

Figure 5 shows the opening pressure (from speaking to breathing mode) as a function of the thickness of the valve's cap (within a single panel) and as a function of the number of holes (between panels). At an increased cap thickness, a clear increase in opening pressure can be observed. The number of air holes seemed to have little effect on the opening pressure. The mixed effect analysis confirmed this. Opening pressure significantly (significance level $\alpha = 0.05$) and linearly depends on cap thickness ($p < 0.001$) and not on the number of holes ($p = 0.64$). The opening pressure range is between 2.9 and 6.0 kPa.

Airflow Resistance Coefficient

Figure 6 shows a comparison of ARCs for different valves (6 iValves, 1 Atos Cassette, and 1 FreeHands) at inhalation, and the physiological upper airway resistance (UAR) at 0.5 L/s.[73] The iValve's ARC graph (fig. 6) in the closed (speaking) mode shows a rapidly declining graph with mean values (±SD) ranging 1533 ± 474 to 270 ± 22, and decreasing with flow.

The iValve's ARC in the open (breathing) mode shows a much flatter graph, with values ranging from 811 ± 63 to 159 ± 17 Pa · s^2/L^2 and slightly decreasing with flow. The ARCs of the Atos Cassette and FreeHands are almost constant and identical. The iValve's ARC in the breathing mode is within the physiological UAR range; in the speaking mode, the ARC is above the UAR and is considerably higher than the ARCs of the Atos valves.

Discussion

The opening pressures and closing flows of the iValve are independently controllable between 2.9 and 6 kPa and 1.8 and 6.0 L/s, respectively. The iValve's ARC is higher than those of the Atos valves, but around 0.5 L/s (quiet breathing), it is just within the upper and lower ranges of the healthy physiological UAR coefficients found in the literature.[73] Reducing the ARC of the iValve in the speaking mode will be one of the targets for the next version in ongoing development.

Closing flow

The iValve's closing flow appears to linearly increase with the number of air holes, and appears to be independent on the valve's thickness (fig. 4). The closing flows of the iValve should be at least above those used in moderately intensive breathing (moderate exercise or daily activities) and below the laryngectomy patient's maximum. Quiet breathing, in laryngectomy patients and normal subjects (alike), is 0.1 L/s.[74] The peak inspiratory flow is 6.0 L/s for healthy male subjects,[75] and 2.5 L/s for asthma patients.[76] Geertsema et al. described an operating range of 1.7 to 3.8 L/s of closing flow in *in vivo* tests of their prototype.[15] These physiologically attainable flows fall perfectly within the range of the iValve.

The closing flow measurements were performed without an HME filter. Arguably, the HME filter will influence (decrease) the closing flow of the iValve. Once the overall ARC has been decreased and HME filter is available, the final prototype will need readjusting to the patients' range of comfortable control flows. Since the relationship between the geometry and the control characteristics are now known, this is completely feasible.

Opening pressure

The iValve's opening pressure appears to linearly increase with cap thickness, yet it appears to be independent of the number of air holes (fig. 5). The pressure's independence of air holes is not surprising, since at closure there is no flow and thus no influence of the air holes. The opening pressure of the iValve should be just higher than the pressures required for intelligible speech in laryngectomy patients. Geertsema et al. described an operating range of 0.8 to 7.2 kPa for the opening pressure in *in vivo* tests of their prototype.[15] Roxburgh and Perry

found a range of 1.52 to 7.85 kPa for speech at normal loudness using the Blom-Singer ATV.[77] Hamadé et al. found a range of 1.96 to 3.92 kPa with manual occlusion.[78] Grolman et al. found a mean pressure of 3.2 kPa at manual occlusion during normal speech, and 6.6 kPa during loud speech.[79] The physiologically attainable opening pressures also fall perfectly within the range of the iValve. The opening pressure measurements have been performed without HME filter. However, no influence on the opening pressure is expected, since the HME filter is positioned past the closed cap.

Airflow Resistance Coefficient

The iValve's ARC decreases with increasing flow, and is not a single ARC. This finding is in concordance with definition of ARC of Verkerke et al.[72] only if we assume the circumferential flap valve progressively opens at higher flows.

The iValve's ARC in the breathing mode is comparable to that of the healthy physiological system. The ARCs of the Atos Cassette and the Atos FreeHands are almost identical and are considerably lower than that of the iValve.

Although an ARC comparable to the normal physiological situation will benefit the pulmonary system on the long-term,[80] patients prefer a much lower ARC. The mean nasal airway (nasal passage, pharynx and larynx) resistance at inhalation ranges from 304 to 742 $Pa \cdot s^2/L^2$ in normal healthy subjects.[73]

The iValve's ARC is overall considerably higher with the iValve in the closed (speaking) mode than in the open (breathing) mode, because in open position the central conduit is open to airflow as well.

Whether a high ARC during inhalation at speech is considered uncomfortable by the patient is unknown and can only be investigated in an *in vivo* study. Values in the healthy UAR lower range are the aim for our next prototype.

Clinical implications

Because all patients are different, the iValve's required opening pressure (which also defines the maximum automatic speech loudness) can be selected by choosing an iValve version from a limited range of products. The iValve's closing flow can be adjusted in each iValve version by adjusting the number of open air holes in the cap.

The ARC of the iValve in the breathing mode already lies within the natural and healthy UAR range, but is higher than the ARCs of both Atos valves. The ARC at inhalation in the speaking mode is above the UAR and an *in vivo* study will have to show whether this is acceptable or will have to be reduced. An ARC that is too high can be uncomfortable to the patient, one that is too low might decrease pulmonary function in the long term.[80]

Conclusions

The iValve works as designed. It opens and closes at the appropriate physiological pressure and flow values. The opening and closing values can be independently controlled, so individual adjustment of the iValve is feasible and simple to implement. The closing flow in this

prototype is adjusted by varying the number of holes in the cap, and the opening pressure can be selected from a limited range of (e.g. 3 to 5) versions.

The airflow resistance of the iValve at low flows and with an HME filter present is higher than that of commercially available valves, but is within physiological values (at 0.5 l/s and higher flows). Only in the speaking mode will the airflow resistance have to be reduced to be comparable to that of commercially available valves. This study shows that the iValve prototype is ready for evaluation in a clinical study.
The iValve's design also shows that using skin-soft materials for the construction of this type of medical device is very feasible and can dramatically decrease the number of assembly parts and consequently potentially decrease manufacturing costs, opening the way to possible disposable solutions, and reaching patients in developing countries. The iValve promises to be a viable asset to the existing range of ASVs, with the added benefit of improved intuitive control and greater voice dynamics for the patient. The *in vivo* tests will show how these characteristics are perceived by patients.

6

In vivo evaluation of the iValve v1

In Vivo Test of a New Handsfree Tracheostoma Inhalation Valve.

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Submitted

Abstract

Objectives: Current automatic tracheostoma valves have the disadvantage that exhaled breathing air is expended on closure of the tracheostoma valve, leaving less air for phonation. Another disadvantage is that the subsequent inhalation opens the valve again; as a result constant phonation is not achievable. To challenge these problems a new tracheostoma valve has been developed that is closed by fast inhalation in contrast to exhalation.

Materials and Methods: In this study a prototype inhalation valve was tested on patients and compared to the current most widely used exhalation tracheostoma valve, the Atos Provox FreeHands HME.

Results: The new inhalation valve was comparable in performance to the Atos Provox FreeHands HME, despite the prototype's high inhalation airflow resistance. An added advantage was that patients were able to whisper using the inhalation valve.

Conclusions: As proof of concept, the inhalation valve has demonstrated to be a viable design.

Introduction

Tracheo-esophageal speech is currently the most widely used technique for voice rehabilitation after total laryngectomy.[81,82] With this technique, an artificial fistula is created in the wall between the trachea and esophagus. A one-way valve prosthesis (voice prosthesis or shunt valve) is placed in this artificial fistula, permitting air from the lungs to pass into the esophagus. To make use of the shunt valve the stoma requires closure with a finger or thumb. However, this is deemed unhygienic and some people cannot securely close the stoma by this means. Furthermore, it draws attention to the laryngectomy status and conversation is hampered if both hands are occupied.
Handsfree tracheoesophageal speech adds to a more natural way of speaking after total laryngectomy. Several tracheostoma valves have been designed in the past, based on the mechanism of closure at exhalation, with the aim of realizing handsfree speaking.[6,11,12] The most commonly used exhalation speech valve is the Atos-Provox FreeHands adjustable tracheostoma valve (fig. 1). This valve is closed by fast exhalation. With every inhalation the valve opens again, therefore continuation of speech after inhaling requires repeating closure of the valve by fast exhalation.

Therefore, this concept has its disadvantages. The most important being that exhaled breathing air is spent for closure of the tracheostoma valve, leaving less air for phonation.[14] Another disadvantage is that the next inhalation opens the valve again, thus constant phonation is not possible.
To challenge these problems, Geertsema *et al*. designed a new mechanism based on the principal of inhalation.[13-15] According to the author, major improvements of this mechanism compared to exhalation valves are the possibility of continuous speech, coughing, and the fact that no exhaled air is spent to close the valve. To incorporate this mechanism so that patients can use it, a new tracheostoma valve has been developed, which is closed by fast inhalation, saving air for phonation during exhalation. The new prototype valve which closes by means of inhalation will be referred to as iValve, whereas the commonly used Atos-Provox FreeHands HME, which closes by means of exhalation, will be referred to as the FreeHands.

The iValve opens at strong positive air pressure (fast exhalation) and closes at strong negative air pressure, at high flow (fast inhalation). This feature enables the patient to switch the valve harmoniously between the “speaking” and “breathing” setting.
Furthermore, the iValve incorporates a second valve, which enables the patient to inhale even when the valve is in the closed “talk” setting, enabling continuous speech.
The characteristics of the iValve can be adjusted by adjusting the number or size of the breathing holes in the cap. Technical specifications and more in depth clarification of the *in vitro* testing of the iValve can be found in: *In Vitro Evaluation of the iValve: A Novel Hands-Free Speech Valve.*[65] As final proof, this study tested the inhalation valve on laryngectomy patients.

Figure 1. The Atos Provox FreeHands HME exhalation tracheostoma valve with a HME filter (left) and a close-up of a patient wearing this valve, connected to a peristomal patch (an adhesive flange that is attached to the skin around the stoma on which a speech valve can be attached) (right).

Aims of the study

The overall aim of the study was to compare speech using the iValve prototype to speech using the commercially available Atos speech valve. Comparison was based upon the maximum phonation time and the maximum phonation loudness. The hypothesis was that it would be possible to speak longer as well as louder using the new iValve as compared to the FreeHands, at equal or lower intra-tracheal pressure.

Materials and Methods

Patients

In total, 26 laryngectomy patients of the University Medical Center Groningen participated in this study. Exclusion criteria were either low lung capacity, for which speech valve operation was deemed too difficult, or patients with irritated or irregular stomas so that the valve could not be attached to the stoma. Only patients who could speak with both iValve and FreeHands were included in the study.

The patient group had a mean age of 64 years (range 45 to 83), most of them male (three females). Of the total population of 26, 14 patients could talk with both valves. The tracheo-esophageal shunt valve used by the study group were; Atos Provox II shunt valve (11 patients), Groningen Ultra Low Resistance (GULR) valve (10 patients), Atos Provox Activalve (4 patients) and one patient did not use a shunt valve.

Recruited patients were scheduled for routine check-up (e.g., shunt valve changes) or came to the hospital only for this study. Measurements took about 75 minutes per patient. All patients signed an informed consent prior to the study.

In total 14 patients were included in the clinical study (fig. 3).

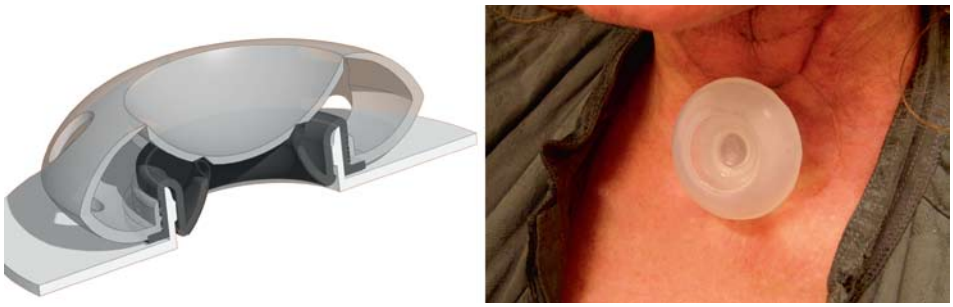


Figure 2. A cross-sectional view of the inhalation valve in closed position (left) and attached to the stoma using an adhesive patch (right).

Valves

In this study two valves were compared to each other:

- The new iValve: There was no dedicated HMEfilter incorporated in this valve, because the type of filter that could be combined with this valve was not available at the time. A nylon mesh was incorporated between the valve and the standard adhesive stoma patch to replace the filter function of the HME filter.
- The commercially available exhalation tracheostoma speech valve: Atos Provox FreeHands HME (manufactured by Atos Medical AB, Sweden) type blue (medium resistance).

All patients were fitted with both valves, which were connected to a standard peristomal flange (Atos Provox FlexiDerm Adhesive base plate Oval).

The patients were randomly divided into two groups of 7 patients each. Group 1 first used the FreeHands, then the iValve; group 2 first used the iValve, then the FreeHands (fig. 3).

At the beginning of the study, all patients received a short training in how to use the speech valves. When they could speak comfortably and when spontaneous speech was possible was the clinical study started.

Voice and speech assessment

During the study the following parameters were measured:

- Loudness and dynamic range in dB: the patient phonated a sustained vowel /a/ at three levels of phonation: as soft as possible, at comfortable loudness, and as loud as possible. After that, the patient had to read a couple of words and sentences, also at three levels of phonation. The sound was recorded, measured and visually plotted in a phonetogram.
- Maximum phonation time in seconds: the patient would phonate a sustained vowel */a/*, */e/* and */u/* at comfortable pitch and loudness after maximum inhalation. Time was measured using a stopwatch. Measurements were repeated three times for statistical analysis (median out of three calculated).

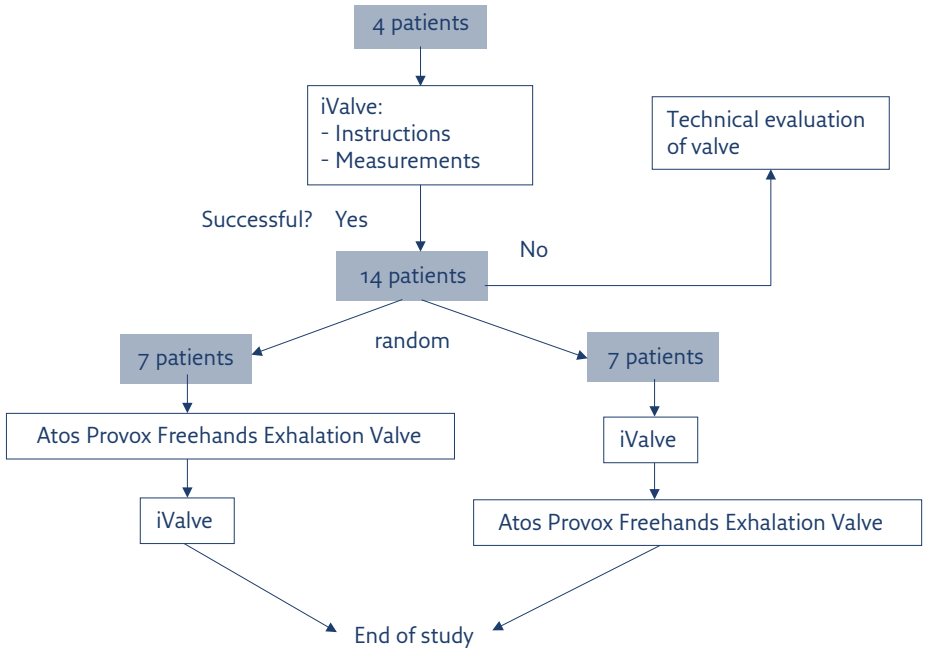


Figure 3. Flow diagram of the study procedure.

- Intratracheal pressure in kPa and flow in L/s: The patient phonated a sustained */a/* at two levels of phonation: as soft and loud as possible. After this, the patients were instructed to speak a Dutch sentence ("Ik heb zin in een kopje koffie", Dutch for "I fancy a cup of coffee"), also at the two levels of phonation. The intratracheal pressure and flow were measured in the test setup as stated in figure 4. Repeated three times for statistical analysis (median out of three calculated).

Structured Questionnaire

After measurement of both valves, the patients were asked to fill in a questionnaire about items regarding the experience when using the two different valves.

To verify the hypotheses, loudness, phonation time and intra-tracheal pressures during speech of all 14 patients were compared using both valves. After this, the values were compared to the opinions of the patients that resulted from the questionnaire.

Statistical analysis

All parameters were analyzed using SPSS version 16.0. Statistical analysis was mainly based on comparison of the two groups. The normality of all parameters was checked visually using histograms and Q-Q plots and, where needed, the parameters were $\log^{10}$-transformed to achieve normality. In case of normality, equality of means between two paired groups was

tested using the paired sample t-tests. In case of non-normality, parameters with two-way classifications were tested using a Mann-Whitney non-parametric rank test. A two-tailed p-value of <0.05 was taken to indicate statistical significance.
To trace confounding variables, linear regression analyses were performed. To test the normality of the residues, a Kolmogorov-Smirnov test was used.

Results

Loudness of speech and whispering
To measure the loudness of the voice (in dB), a phonetogram was composed. This was carried out by means of a loud and a soft sentence using both valves. Using these two variables, the dynamic range was calculated (loud minus soft). An important finding was that patients were able to whisper using the iValve as the valve doesn't need to be held in closed position during exhalation. Using the FreeHands, whispering was not possible.
The results of the measurements of loudness are expressed in table 1 and figure 4.

In the questionnaire, all patients were asked what they experienced to the loudness of their voice when speaking with the iValve. 44.4% experienced a softer voice using the exhalation valve, 44.4% a louder voice and 11.1% could not point out a difference.
When the results for loudness between both valves are compared, there is no significant difference found between either a loud sentence (p = 0.068), a soft sentence* (p = 0.394) and the dynamic range (p = 0.744).

Phonation time
The results of the measurements of phonation time are expressed in table 2 and figure 6.
Comparison of iValve with FreeHands with regard to phonation time showed no significant differences during the vowel */a/** (p = 0.713), */e/* (p = 0.844) or */u/** (0.836).

Length of sentences
Another item in the questionnaire was the length of the sentences when using the iValve as compared to the FreeHands. 27.8% experienced a louder voice using the FreeHands, 27.8% a softer voice and 44.4% were not aware of a difference.
When the results for phonation time are compared between both valves, there is no significant difference when using the vowel */a/* (p = 0.713)*, */e/* (p = 0.844), */u/* (p = 0.836)* which is in accordance to the opinion of 27.8% of the patients.

Intra-tracheal pressure
The results of the intra-tracheal pressure measurements during speech are presented in table 3 and figure 7.
Regarding intra-tracheal pressures there are many outliers, indicating large differences between patients.

*not normally distributed.

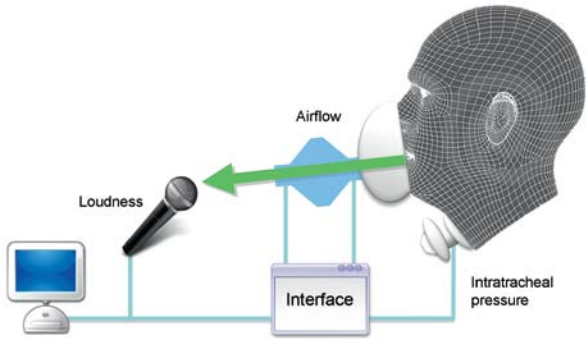


Figure 4. The test setup used to measure the flow, intra-tracheal pressure and loudness of voice during phonation.

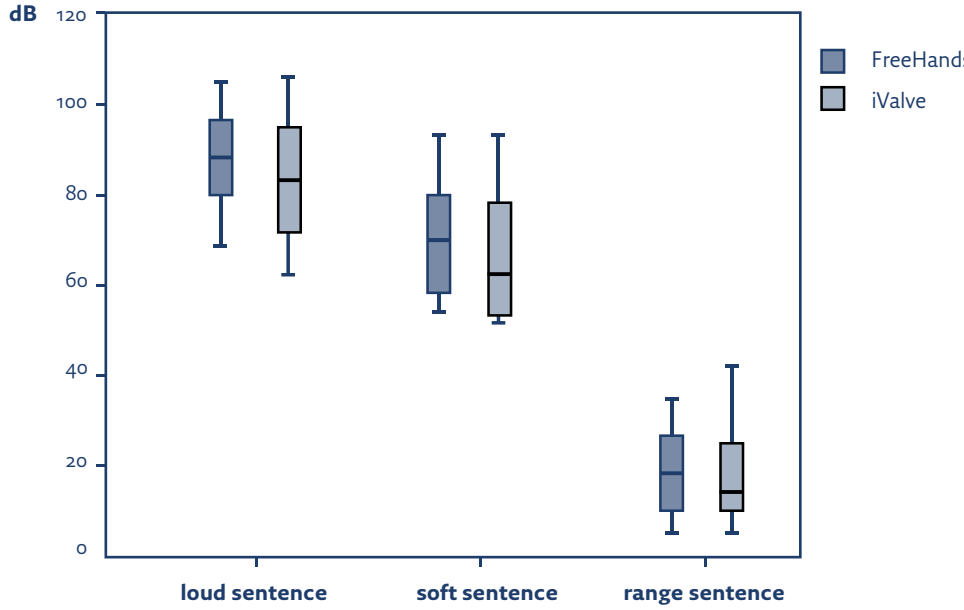


Figure 5. Loudness in dB of the inhalation valve as compared to the FreeHands exhalation valve when using a loud and a soft sentence.

Table 1. Measurement results for loudness of a loud sentence, a soft sentence and the range (loud minus soft) using the Atos Freehands and iValve speech valves.

Valve		Valid n	Min.	25%	Mean	SD	Median	75%	Max.	Var.
FreeHands	Loud sentence	14	68.0	79.5	88.1	9.9	89.0	96.25	103.0	97.5
	Soft sentence	14	55	59.0	69.9	11.7	71.5	81.0	90.0	136.8
	Range	14	6.0	10.8	18.1	8.5	18.0	26.5	32.0	72.6
iValve	Loud sentence	14	63.0	72.5	84.8	12.3	85.5	95.3	104.0	151.3
	Soft sentence	14	55.0	56.0	67.3	12.4	62.5	78.3	90.0	153.1
	Range	14	6.0	10.25	17.5	9.6	14.0	24.8	40.0	91.5

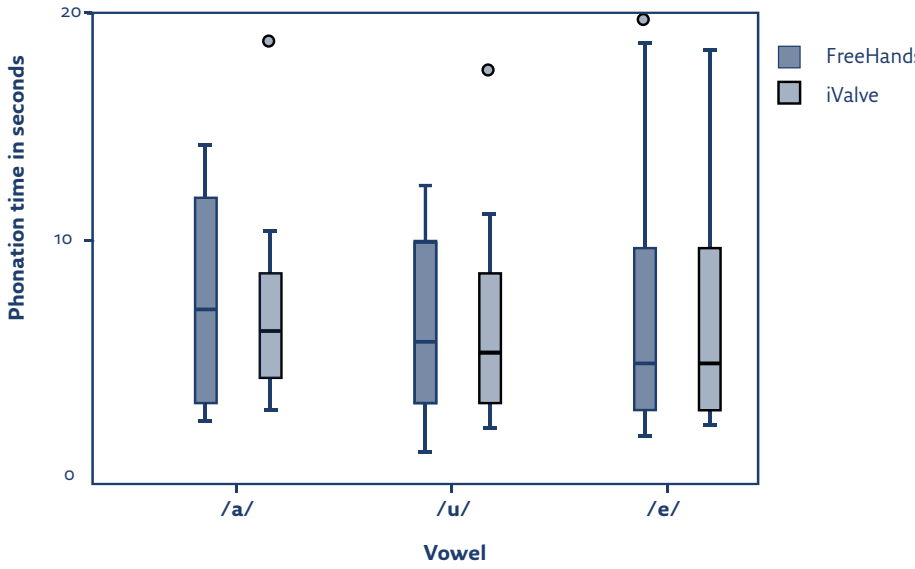


Figure 6. Comparison of phonation time between iValve and FreeHands in seconds.

When the results for intra-tracheal pressure are compared between both valves, there is no significant difference when expressing a loud vowel */a/* (p = 0.708), soft */a/* (p = 0.158), range */a/**(p = 0.341), loud sentence* (p = 0.168), soft sentence* (p = 0.089) and range sentence.

Results of the questionnaire

63.6% of the patients have the opinion that closing the stoma with a finger or thumb is not annoying, because they are used to it. However, 85.7% think handsfree speech makes life easier, as they can speak while both hands are occupied. All patients would use an automatic speech valve, if it would work optimally and usage is not difficult. 33.3% of the patients are of the opinion that usage of the iValve is more difficult than closing the stoma with the hand. If the iValve and FreeHands are compared according to usage, 57% think the iValve is more easy to use.

When questions are asked about the flaws of the iValve, 81% is of the opinion that inhalation in closed position is too difficult, 47.4% think the valve opens too easily with loud speech, and 54.5% think that the whistling noise when the valve opens is annoying.

*not normally distributed.

Table 2. Measurement results of phonation time (in seconds) using the Atos Freehands and iValve speech valves.

Valve	Phonation time in s	Min.	25%	Mean	SD	Median	75%	Max.	Valid n	Var.
FreeHands	/a/	2.3	3.0	8.1	6.3	6.7	11.3	25.3	14	39.6
	/u/	1.0	3.0	7.4	6.6	5.3	10.2	25.3	14	42.9
	/e/	1.7	2.5	7.0	5.8	4.5	9.6	20.0	14	34.0
iValve	/a/	2.7	3.9	6.7	4.2	5.8	8.3	19.0	14	17.8
	/u/	2.0	2.9	6.2	4.3	5.0	8.3	17.7	14	18.1
	/e/	2.3	2.7	6.4	4.4	4.5	9.1	16.7	14	19.7

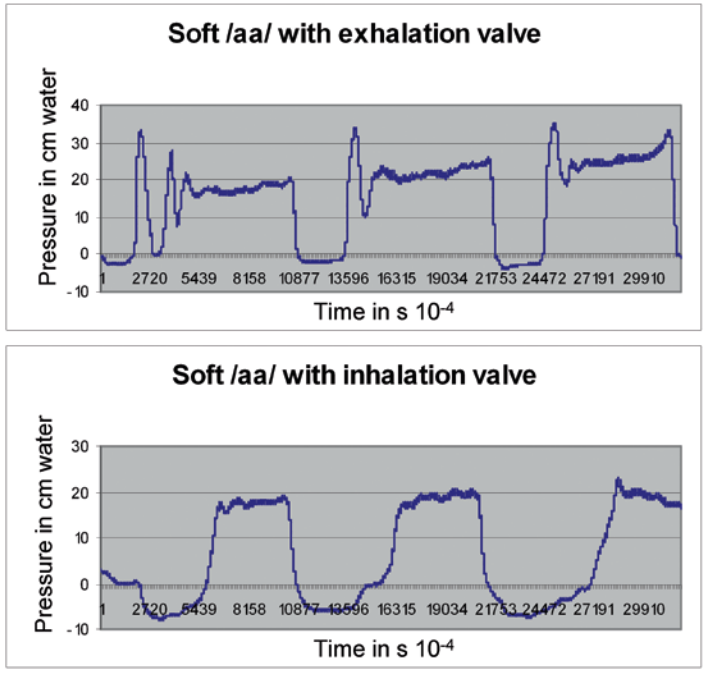


Figure 7. Intra-tracheal pressure during speech using an FreeHands exhalation (left) and an iValve inhalation valve (right). A clear peak pressure was measured during closure of the FreeHands exerting stress between the stoma and stoma-connector.

Discussion

The absence requirement to close the new inhalation valve using a fast exhalation resulted in new possibilities. The most notable characteristic of using the new iValve proved to be the ability to whisper, which was not possible using the current exhalation valves. This, however, could not be registered in a reliable way using our phonograph, which mainly responds to a fundamental frequency. Therefore it cannot be derived from the above-mentioned results, but only from our observations.

The patients' preference regarding a specific type of valve could be caused by their needs regarding the supplied air resistance of the speech valve. However, the aerodynamic properties or the iValve were not adapted to the patients' needs during this study. The final results could be optimized if the speech valves were individually tailored to the patients.

When patients were asked which valve was least difficult to use, 42.9% of the patients said they noted no difference between valves. In general, 23.8% preferred the iValve and 33.3% the FreeHands. A difference between the valves was that the FreeHands was tested with an HME filter and the iValve without. This could account for a difference in resistance through which the iValve is more easy to use. Because the majority of the people detected no difference between the valves, this effect was not taken into account.

Table 3. Measurement results of intra-tracheal pressure of a loud sentence, a soft sentence and the range (loud minus soft) using the Atos Freehands and iValve speech valves.

Valve		Valid n	Mean	SD	Var.	25%	Median	75%	Min.	Max.
FreeHands	Loud /a/	14	6.5	3.4	11.7	4.6	5.6	7.9	2.4	16.6
	Soft /a/	13	3.6	1.8	3.1	2.8	3.4	4.7	0.3	6.7
	Loud sentence	14	7.4	4.1	17.1	5.3	5.7	8.7	2.7	18.5
	Soft sentence	14	4.1	2.0	4.0	2.9	3.7	5.0	0.5	7.6
	Range /a/	14	3.1	3.1	9.4	0.5	3.1	5.4	-2.0	9.9
	Range sentence	14	3.4	3.3	10.6	1.4	2.0	4,1	0.9	10.9
iValve	Loud /a/	14	6.2	3.1	9.9	4.4	5.5	7.2	2.2	13.4
	Soft /a/	13	2.8	0.8	0.6	2.1	2.8	3.7	1.7	4.0
	Loud sentence	14	5.9	2.2	4.7	4.8	5.3	6.6	3.7	12.7
	Soft sentence	13	2.9	1.1	1.2	1.8	3.1	3.8	1.2	4.8
	Range /a/	15	3.4	3.0	8.9	1.9	2.6	3.3	0.5	10.7
	Range sentence	15	2.6	2.5	6.3	1.2	2.3	2.6	-0.9	9.6

A challenge to the current prototype version of the iValve is that a majority of patients are of the opinion that the iValve's dimensions are too big. Furthermore, 81.8% of the patients point out that inhalation in closed position is too difficult and insufficient. These challenges have to be met to obtain an optimal product. However, it must be noted that iValve study is still a feasibility study, as proof of concept. If the challenges as mentioned above can be addressed, then the iValve concept could be even more promising since patients can already speak as well with the FreeHands as with the current prototype iValve.
Therefore, more possibilities have been generated to improve quality of life for laryngectomy patients. The missing link in this domain is the fixation of speech valves. They have to be connected to an intra-tracheal tube or to peristomal flanges or patches. Poor fit of these patches to the stoma, caused by variations in stoma geometry, is the biggest issue. Of all participants in this study, 50% indicate that patches do not stay attached for longer than a few hours. An automatic speech valve causes more stress on the patch than manual closure of the stoma. To increase the application of automatic speech valves, the issue of their fixation has to be solved.

Conclusion

Based on aforementioned results and discussion, the new iValve inhalation speech valve proves to be comparable to the existing speech valve. The foremost characteristic of the iValve is the ability to allow whispering, which is not possible using the current exhalation valves. However, further engineering has to be performed to optimize the new inhalation valve. Especially ease of inhalation in speaking mode will have to be improved upon. Still, the new iValve has proven to be a very viable, low cost design. With the air resistance addressed, the iValve is an optimized product, which could surpass the efficiency and ease of use of current automatic tracheostoma valves.

7

Recent developments

Recent developments

The studies in this thesis, in particular Chapters 4, 5 and 6, treat the development and testing of the iValve v1, a handsfree speech valve. In this chapter, we present other prototypes that were developed during this project, but were not covered in the studies presented in the previous chapters. The iValve v2 presented here is a next, and improved version of the iValve v1. The iPatch v1 and v2 contoured adhesive patches are essential for the goals of this project, are currently under testing but have not been presented before in this thesis.

Improved adhesive patches

Chapters 2 and 3 show that the major contributor to the low use of automatic speech valves, apart from the drawbacks of current handsfree systems, is the poor fit and poor adhesion to the peristoma of the required adhesive patches. Figure 1 shows a typical tracheostoma, and a regular patch (Atos Provox Optiderm oval) that the patient improved herself to make it fit better. This patch, although now good enough for use with the manually occludable valves, is, according to the wearer, still not good enough for durable (day-long) use with a handsfree valve. This is because handsfree valves exert considerably higher stresses on the adhesive patch as the speaking pressure of the valve is not countered by a pressing finger, which is the case in the manual occludable systems. Even the much lower peak pressures of the iValve do not allow this patient to use handsfree speech for over a prolonged time.

Our patient, like the majority of patients, needs a better fitting adhesive patch in order to be able to use a handsfree speech valve. That is why we also developed improved adhesive patches alongside the development of improved handsfree speech valves. Our goal was a new adhesive patch that would outperform the commercially available adhesive patches both in attachment time and comfort: a patch that attached so well, that day-long handsfree speaking would be possible.

The results of patient testing, however, were not available at the time of printing of this thesis: only preliminary results of tests on single patients will be presented here.

The iPatch contoured adhesive patches

The study presented in Chapter 2 delivered data on the stoma shape of 191 laryngectomy patients. For the development of an improved stoma adhesive patch it was crucial to know the difference between the peristoma shape of those patients who can, and those who cannot use commercially available adhesive patches. We found this difference. This data (presented in table 3 of Chapter 2) was used to develop the iPatch adhesive patch (fig. 2). This design has some improvements over the most widely used adhesive patch, the Atos Provox Flexiderm Oval and Atos Provox Optiderm Oval adhesive patches (Atos Medical, Sweden).

First of all, the shape of the patch was changed from the traditional flat patch shape to a more concave shape (fig. 3). The concavity used was based upon the average stoma depth of 18 mm and a fold-to-fold distance of 34 mm of non-patch users from table 3 (Chapter 2). The patch was designed a bit wider than the required fold-to-fold distance to provide some extra perpendicular force to the adhesion area, using the high elasticity of the selected patch

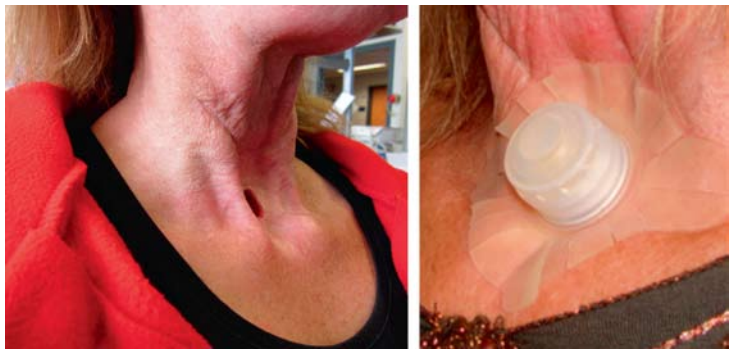


Figure 1. Typical tracheostoma (left), and a regular patch (Atos Provox Optiderm oval) that the patient improved herself to make it fit better (right).

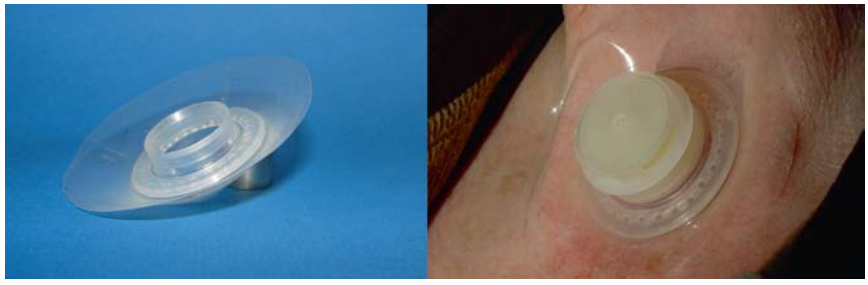


Figure 2. The iPatch version 1 concave adhesive patch, and on a patient.

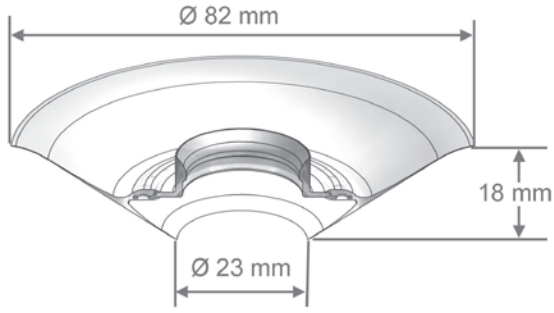


Figure 3. Cross-section of iPatch v1 rendering with dimensions. These dimensions were based upon the measurement data from the stoma geometry studies of Chapters 2 and 3.

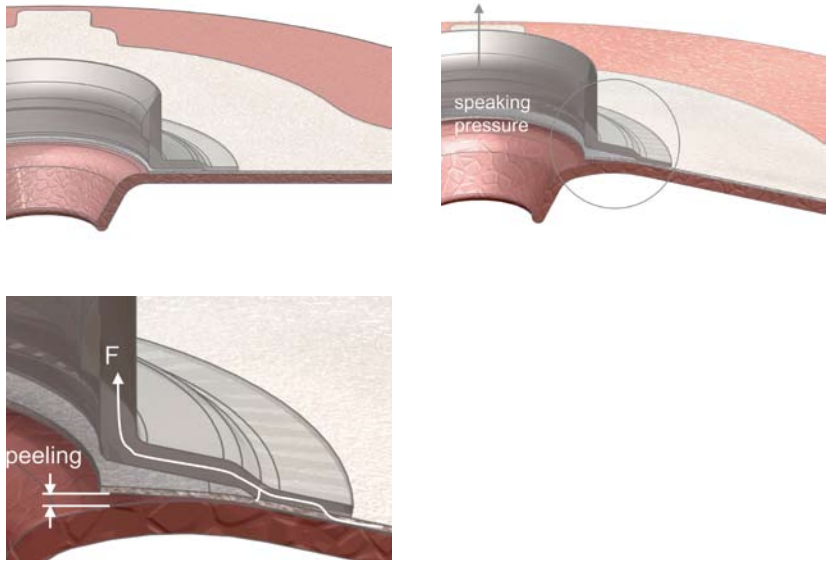
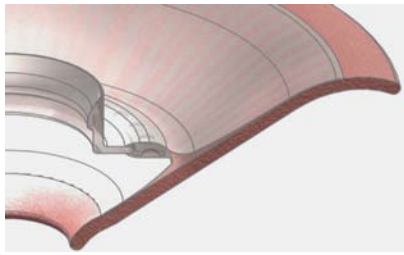
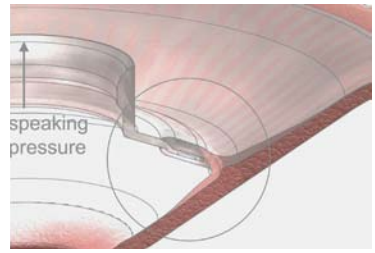


Figure 4. Cross-section of a traditional patch on skin. Initially no pressure is exerted on the patch (left). When speaking pressure is applied (right), the stress is conducted through to the adhesive material that, due to its stiffness on the stoma side of the connector, is lifted up in its entirety, thus promoting peeling (close-up).

material. The P95 stoma diameter of 25 mm of the entire population from table 2 (Chapter 2) was taken as a reference for the central opening. Some extra space was introduced to make centering the patch more undemanding.

The patch material in contact with the peristomal skin was designed to be extremely thin and elastic, allowing movement of skin at changing postures. Due to this elasticity, thinness and transparency, it is barely visible on the skin. This was indeed reported by some patients to be very comfortable: almost unnoticeable, even under strain.
Last, we addressed the failing behavior of the traditional patch. We observed that in most traditional adhesive patches, when they fail under the phonation pressure exerted on the patch at handsfree speech, failure starts at the inner rim of the adhesive patch: a process called *peeling*. This is due to the strain on the patch, perpendicular to the patch surface, close to the edge of the patch. In most commercially available adhesive patches this failure mode inevitably occurs after some time when the relative stiff patch on the stoma side of the connector is lifted up in his entirety by the puling stress on the connector (fig. 4 left). On the distal side of the connector, the stress on the connector is conducted through the patch material parallel to the skin, as a shear force (fig. 4 right and close-up). The adhesive is much stronger when under shear force then under pure strain. This failure mode is prevented in the iPatch design: the connector (identical to the connector used in the Atos Provox patches)





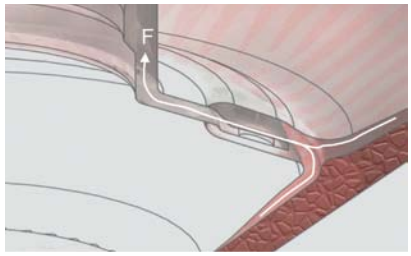


Figure 5. In the iPatch design, the connector is connected to the adhesive patch part of the valve almost halfway the diameter of the patch (left). Stress on the connector is now conducted to the patch material parallel to the skin and converted into a shear force on the adhesive agent, distributed over the whole of the area: no peeling occurs (right and close-up).

Figure 6. The iPatch v1 (foreground) is molded in a three part PMMA mold (background). The connector used in the iPatch is perforated around the perimeter (visible in the iPatch) and inserted into the mold prior to filling.

is connected to the adhesive patch part of the valve almost halfway the diameter of the patch (fig. 5 left). Stress on the connector is now conducted to the patch material parallel to the skin and converted into a shear force on the adhesive agent, distributed over the whole of the area (fig. 5 right and close-up).
Both iPatch v1 and v2 were manufactured using the same silicone rubber as all other devices of this thesis (Zoo1, "Bellows" platinum cured silicone rubber, Principality Medical Limited, Rogerstone, UK). This two-component silicone rubber is injected into a pre-cooled mold (fig. 6), in which a standard Atos Provox connector (courtesy of Atos Medical, Sweden) is inserted prior to closure and filling of the mold. The connector is first perforated around its perimeter to facilitate integration with the silicone part. The mold is filled using a cartridge gun with mixing nozzle. The mold is then placed inside a small pressure vessel, pressurized at 7 bar to prevent air bubbles in the material, and cured in a conventional convection oven at 70°C for about 20 minutes. The silicone rubber forms a form fit around the Atos connector, so that once taken from the mold, the connector and silicone patch form one single part.

The adhesive selected for the first prototypes was the adhesive that was also used on the Atos Provox FlexiDerm Oval patches. Identical adhesives were selected so that in a later stage both patches could be compared in a single study, varying only the patch shape.
Since the silicone rubber surface of the iPatch is apolar and very hydrophobic, the adhesive does not naturally bond to it: the surface requires preparation first. A plasma treatment was used to prepare the silicone rubber surface for binding: a high voltage electric plasma is generated over the silicone surface, which causes the silicone polymer chains to break and activate. The increase in the surface free energy facilitates better binding with the adhesive. This is a standard treatment for materials that naturally adhere very poorly to adhesives.
Figure 7 shows the fit of the iPatch v1 to the peristoma of a patient during a patient test. Especially noticeable is the conformability of the patch at different (extreme) postures of the neck, where the patch follows all pleats and folds of the skin. The transparency of the material renders the patch almost invisible. Remarks from some patients were either that they liked the iPatch being almost invisible or that they disliked the fact that it made the skin look shiny. All patients so far thought the iPatch was very comfortable and almost unnoticeable.
The full outcome of the patient test was unavailable at printing of this thesis.

An early pilot test of the iValve on two patients showed that minor optimizations in the shape design could even further improve the fit of the iPatch. From these observations, the drop-shaped iPatch v2 evolved (fig. 8). At test-fitting on patients (fig. 9), this iPatch v2 seemed to perform very well, but a formal patient study still has to prove this.
Although no study data were available at this point, an adhesive patch that fits better to tracheostomas than commercial adhesive patches seems possible without surgical intervention or expensive tailor-made molded solutions.

The iValve v2 hands-free speech valve
In Chapter 5 we have shown that the iValve version 1 works well: our implementation of the inhalation principle performs within physiological boundaries and the valve performs within

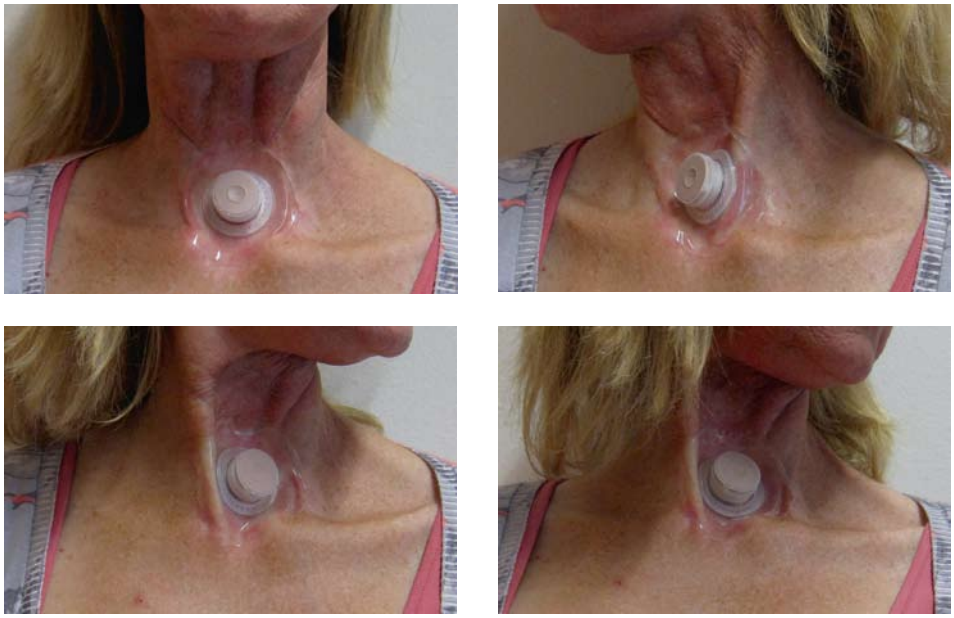


Figure 7. The iPatch v1 on a patient during a patient test. Clearly visible is the conformability of the patch at different (extreme) postures of the neck, where the patch follows all creases and folds of the skin.

Figure 8. The drop-shaped iPatch v2 (foreground) and its mold (background).

the specification of the relatively complex prototype of Geertsema et al.[14,15] Our patient study of Chapter 5 shows that the iValve works on average just as well as the commercial, and mature, Atos Provox FreeHands valve, and that it is very well perceived by patients: preferred even, over the Atos, if some shortcomings were to be addressed (see Chapter 5). Points for improvements were the high inhalation airflow resistance, the overall size of the valve and the noise during inhalation.
Meetings with commercial partners also revealed that the use or integration of a standard, commercial, HME filter was preferred, instead of the proprietary filter we used in our version 1 prototype.
These points were implemented into the design of the iValve version 2 (fig. 10), of which the development was started alongside the iValve v1. The iValve v2 is based upon the same bi-stable buckling principle as v1 in terms of closing and opening, and therefore is designed to have the same speech characteristics as the v1. The redesign reduced the number of required assembly parts even more, securing a lower manufacturing cost. To give an idea of the part reduction: the Atos Provox FreeHands valve comprises 13 separate assembled parts, excluding the HME filter, the iValve v1 consists of 2 parts and the iValve v1 reduced the number of parts to one. This redesign also simplified the required molds: figure 11 shows the iValve v1 and its molds and figure 12 the mold of the iValve v2.
The function is technically the same: in outward buckled shape, all breathing air is led through the air holes (fig. 13) and through the HME filter: the valve is in breathing mode. At strong inhalation, the pressure under the cap decreases to the point where the cap buckles inward. At buckling inward the attached valve skirt unfurls outward, and drapes itself over the air holes. At exhalation this skirt is pressed firmly against the air holes, blocking all flow: the valve is in speaking mode, and all exhaled air is routed through the voice prosthesis voicing the patient's speech. At inhalation, the skirt is sucked in, a little, allowing air to flow inward: inhaling in speaking mode is thus possible. At strong exhalation, the air pressure under the cap increases until the cap buckles outward, back to breathing mode. Crucial to the flawless working of the iValve is that at the unfurling of the skirt valve, the skirt's movement at the end of its path is parallel to the surface of the air holes: only then will the skirt cover the air holes without folds or crevices for a perfect air tight seal.
Normally a shape like this, with its retracted and furled, extremely thin skirt, would be impossible to manufacture out of a single piece of silicone rubber: the skirt in its relaxed stated presses against the air hole surface and no mould part can be designed to manufacture that shape. Fortunately rubber, with its high elasticity and large deformability, allows products to be partially molded inside-out. When folded inside-out, the iValve requires a very simple mold for manufacturing. For prototyping a 5-part mold sufficed (fig. 12).

Preliminary *in vitro* testing shows that, at least *in vitro,* the opening and closing flows and pressures are similar to the iValve v1 prototype. Major improvements in the design simplified the manufacturing process to a five-piece mold, reduced the number of parts from two to only one, and reduced the size and shape to that of the Atos Provox FreeHands valve (fig. 14). The iValve v2 now connects to a standard Atos Cassette HME filter. The inhalation air flow resistance was reduced to the level of the Atos Provox FreeHands and Atos Provox Cassette (fig. 15).

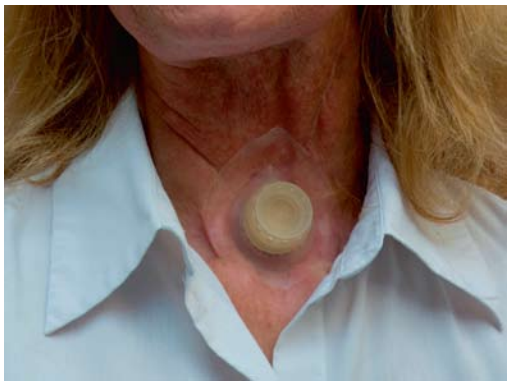


Figure 9. The drop-shaped iPatch v2 on a patient combined with an iValve v2.

Figure 10. The iValve v2, fresh from the mold and folded inside-out (left), with HME filter on an Atos Provox Flexiderm oval patch (middle and right). In breathing mode (middle) and speaking mode (right).

To accommodate the differences in patient physiology and preferences, several designs to make the opening pressure and closing flow of the valve variable, were investigated. On the iValve v1, this option was embodied in a selection from a range of different caps with the appropriate opening pressure, and a means of changing the total area of inhalation holes to change the closing flow. In the v2 prototype, one of these options is the selection of a strength of elastic band that changes both opening pressure and closing flow (fig. 16 shows a prototype where a broad elastic band was used to change the characteristics). One of the elastic band embodiments also incorporates the possibility of changing the total area of air holes by covering/uncovering the air holes.
A preliminary test of the effect of a elastic band on the opening and closing behavior showed a linear relationship between the elastic band strain and the pressures and flows.

The iValve already works well and is only getting better. Currently, efforts are being made to transfer our knowledge to industry and have the iValve and iPatch mature into commercial products.

Figure 11. The iValve v1 two-part design (foreground): cap and capseat (capseat in its "fresh from the mold" state with its valve flap not yet folded inward). Mold (parts) in the background.

Figure 12. The iValve v2 (foreground) in the shape it is molded in, before it is flipped inside-out. The mold is in the background.

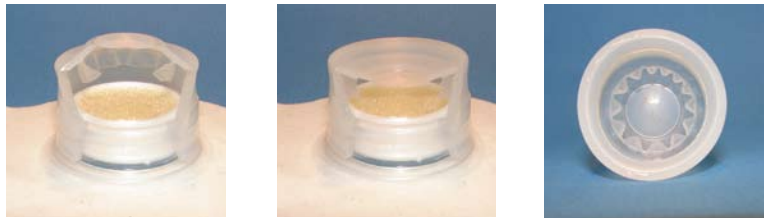


Figure 13. The working of the iValve v2: In speaking mode the cap is buckled outward, retracting the skirt valve to its unfurled state, away from the air holes (left). On the right, the iValve with its cap in buckled-inward shape, its skirt valve furled outward covering the air holes (middle). Right: the iValve viewed from its underside with the skirt valve shown retracted in breathing mode.

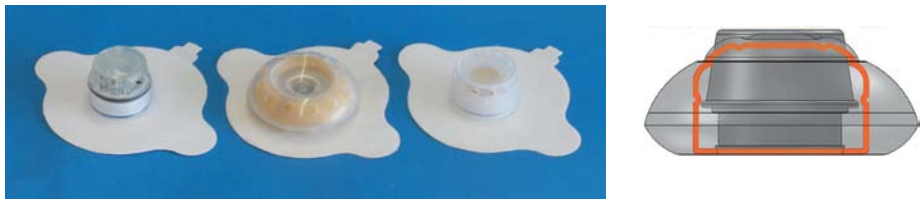


Figure 14. From left to right: the Atos Provox FreeHands The iValve v1 and v2 (left). The Atos provox FreeHands Cassette (dark grey), compared in size against the iValve v1 (light gray) and iValve v2 (orange contour).iValve with its cap in buckled-inward shape, its skirt valve furled outward covering the air holes (middle). Right: the iValve viewed from its underside with the skirt valve shown retracted in breathing mode.

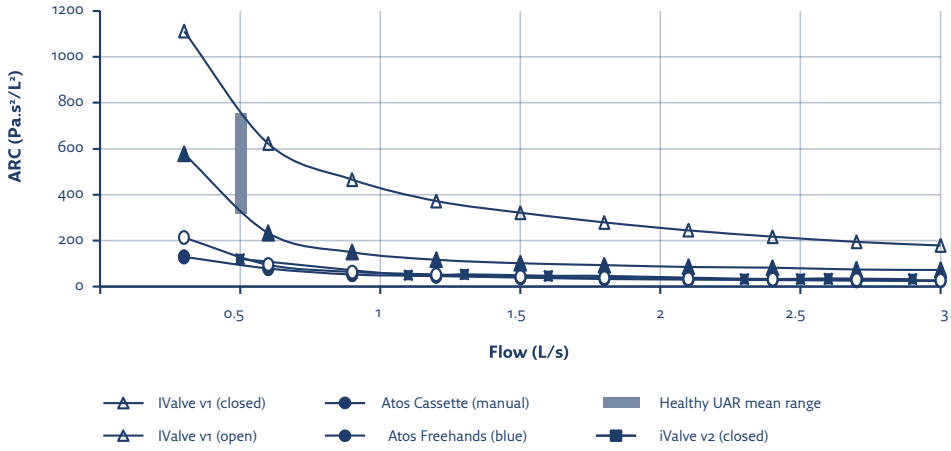


Figure 15. The inhalation airflow resistance coefficient (ARC) of the iValve v1 (in both speaking (closed) and breathing (open) mode), iValve v2 (in speaking (closed) mode) and Atos Provox FreeHands and Atos Provox Cassette: the ARC of the iValve v2 is similar to that of both Atos valves.

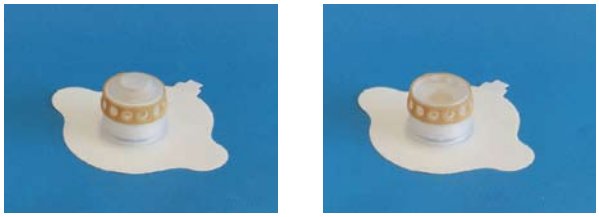


Figure 16. The iValve v2 prototype fitted with a broad elastic band to select the opening pressure and closing flow according to the preference of the wearer. This element could also allow changing the total area of air holes by covering/uncovering the air holes. Breathing mode (left), speaking mode (right).

8

General discussion and conclusions

General discussion and conclusions

This chapter discusses the overall results and achievements of the studies published in this thesis, including the technology of designing in elastic materials that we had to master. Possible business models and financial aspects of the introduction of our new products will be analyzed and discussed in the economic impact section below. Next, in the future developments section we will sketch a roadmap for future developments and trends in health care.

The handsfree speech valve project

Chapters 4, 5, 6 and 7 describe the numerous prototypes that have been manufactured in the course of this thesis: the iValve v1 and v2 handsfree speech valves and the contoured adhesive patches iPatch v1 and v2, as well as all the equipment especially developed to test and manufacture these prototypes. All the studies in this thesis delivered the data that were used to develop these prototypes and aid in future improvements or new developments. All these prototypes, equipment and data together form the "deliverables" of this project. An important deliverable is the "list of requirements and wishes": the list of minimal specifications for our prototypes. Should a prototype not meet its requirements, then it will fail to help us reach our goals. Therefore the requirements are really the specifications of the goals.

We will now analyze the overall succes of the project according to the goals formulated in Chapter 1, in that same order.

The primary goal is crucial to any innovative project: *"to increase the patient's quality of life"*. If this bottom line is not reached which -in this case- means if the patient's quality of life is not increased, the product is not successful. This primary goal has several sub-goals and these are the working goals for the product. But although it is more than obvious: never should the primary goal be forgotten, or the project will not be successful! Often the primary goal is part of the mission statement of the executing company. The first working goal is *"to enable speaking in longer sentences with mid-sentence pauses"*. We translated that goal into several requirements. That was after the decision was made that the handsfree speech valve would be based upon the inhalation principle introduced by Geertsema et al. This is why the requirement states that *"the device operates in two settings: a breathing setting in which the patient can comfortably breathe in and out, and a speaking setting in which the device will block exhalation"*: this is the Geertsema principle of valve operation.

The other requirements originating from this goal are that *"the device allows pausing mid-sentence without loss of phonation air"* and that *"the device provides more air for phonation than commercially available handsfree valves do"*.

In Chapter 6, we tested the iValve v1 handsfree speech valve with regard to meeting these requirements. It appeared that the iValve is a very good handsfree speech valve: it is, in general, *at least as good as* the elegantly designed Atos FreeHands valve, and was actually preferred slightly more often than the Atos valve (57%). According to this study, there was no significant difference in the sentence length between the two valves: here the iValve *is as good as* the Atos. The other advantages of the inhalation principle on which the iValve is based were clearly demonstrated: mid-sentence pausing is now possible, as well as whispering.

The disadvantages that were reported by patients – difficult inhalation through the valve in speaking mode and the larger physical size of the prototype – were taken into account in the development of the iValve version 2, described in the recent development section. Preliminary testing shows that this new prototype, *in vitro*, has the same characteristics as the version 1, but with a notably lower airflow resistance. Thus the requirement that *"the device's airflow resistance in breathing mode is comparable to or lower than that of commercially available handsfree valves…."* seems to be met with the iValve v2. Yet, this requirement is appended with *"…and comparable to the airflow resistance of the healthy airway"*, which was found to be a contradictory requirement since Chapter 5 shows that the natural air flow resistance is significantly higher than the resistance of the Atos Provox valves, and not favored by patients. We decided to make the air flow resistance comparable to the Atos valve: increasing the air flow resistance – should it be required – is then always possible.

The goal *"to increase the use of automatic speech valves and filters in particular, and stoma patches in general, by laryngectomy patients,"* translated into many requirements. Chapters 2 and 3 show that increasing speech-valve use is not only a matter of introducing a better speech valve, but that there is a cascade of causes for the current low handsfree valve use. After discovering the causes we decided to develop both a better speech valve and a better adhesive patch for maximum effect. The requirements have been formulated so that both products can be used individually and in combination with other (commercially available) products.

In case of the iValve, the goal *"to reduce the stress on stoma-interfacing devices"* was translated into the requirement that states that *"the device generates lower tracheal pressures at equal phonatory sound pressure levels…."* so that the (speech pressure) stress exerted on adhesive patches is lower. In Chapter 5 we found that, at equal speaking loudness, the mean intratracheal pressures required during speech are not significantly lower than those of the Atos valve. In most patients however, we did notice a high pressure peak at the start of speech using the Atos speech valve; this peak is absent in the iValve. Since all intratracheal pressure is also exerted on the interfacing device – usually an adhesive patch – a lower speaking pressure will have a beneficial effect on the durability of the interfacing device. Still, patient testing will have to confirm this hypothesis. An added advantage of the iValve is that during use the patient inhales in order to close the valve, so the (negative) inhalation pressure can re-attach the adhesive patch to the skin, comparable to finger pressing on the adhesive patch by the patient to reattach the adhesive patch.

It is still difficult to predict whether or not the iValve will really contribute to increased speech-valve use in patients. After all, this result is not only dependent upon the characteristics of the valve but also upon successful market introduction. Currently we are still in the process of finding a commercial partner, but patient testing so far has shown that the iValve itself is at least as good as the Atos valve, with the advantage over the Atos and other commercial valves of its economical design, which at least theoretically gives the iValve all the right cards for marketability.

The requirements for the iPatch, formulated with the aim to increase the use of speech valves and filters, all focus on at a better fit of the patch to the stoma. Chapter 2 confirms our initial hypothesis of the poor fit of patches: commercial patches show a shape mismatch with the majority of stoma shapes. This study also delivered the data on which to base better fitting adhesive patch shapes. The iPatch version 1 concave shape was based upon an average stoma depth of patients who did not use commercial (flat) adhesive patches of 18 mm and a fold-to-fold size of 34 mm.
The studies presented in Chapters 2 and 3 led to in the development of two improved adhesive patch prototypes, but not in tested products. Preliminary testing shows the iPatch does fit very well to the stoma of patients and is generally perceived very well. Thus our goals *"to realize a better fit and adherence of stoma interfacing devices to the stoma"*, and *"to reduce the geometrical and stiffness mismatch between stoma and stoma interfacing devices"* have not yet been confirmed by thorough patient testing.

And last, we set out to *"match the stiffness and hardness of devices to the surrounding tissues"*. Both the iValve and iPatch of this thesis are made out of silicone rubber. Not only has this material been selected for these prototypes: soft elastic materials such as silicone rubber could be the materials of choice for any device in contact or embedded in soft tissue. In fact, the selected materials are perceived by patients as pleasant to the touch and, in the case of our preliminary findings in patient testing of the iPatch, the patch was perceived to be very comfortable even at extreme head postures. Patients described the patch as almost unnoticeable. Even visually the patch's transparency renders it almost invisible.
In conclusion, we can say that most goals have been reached or that they are on their way to be reached, and that our prototypes have the potential to further increase the patients' quality of life.

Designing in soft materials
By optimally using the properties of elastic materials – beyond the way in which elastic materials are traditionally used – we managed to reduce the number of valve parts from 13 (the commercial Atos Provox FreeHands) to two (the iValve v1) and eventually one (the iValve v2). This incredible reduction of parts can only be achieved if the properties of rubber are optimally used.
The vast majority of design rules in product design are implicitly linked to the conventional and relatively stiff and hard materials such as plastics, metal and wood. Rubber in products is used for its flexibility, but in a very limited way: the majority of all rubber materials feature as sealants, drive belts, hosing and tubing, or liquid or air containers. The property of elasticity is used only to a very limited extent in the design in these examples: these products only exploit the higher elasticity in their function, and not in the way they are manufactured or designed. Elasticity is only a property of the end product in these circumstances, not a tool in the making and designing of the product. If elasticity and softness are used as degrees of freedom in the design or manufacturing process itself, than revolutionary new product shapes and functions can be created: parts can be designed inside-out, inflated, rolled-up or buckled. Such things are impossible in conventional stiff materials. Recently more products

Figure 1. Weener-Ibérica (Weener Plastic Ibérica, S.L., Spain) non-drip valve as can be found in sauce or shampoo dispenser caps (left and middle): when the bottle is squeezed, the little valve (right) moves forward and then allows the cross shaped slits to open, when the pressure drops the cross closes and the little dome retracts, shedding the remaining droplets: the valve cleans itself.

Figure 2. Lékué Cooking Mesh: silicone rubber used as a stretchable cooking basket/mesh (Copyright Lékué S.L., Barcelona, Spain).

Figure 3. A stoma-plug that enables the measurement of tracheostoma pressure during speech in laryngectomy patients. The cushion feature is molded inside-out.

have appeared on the market that exploit the possibilities of elasticity and softness to a further extent: the Spanish "Ibérica" branch of the German Weener packaging company (Weener Plastic Ibérica, S.L., Spain) developed a "non-drip" valve that is very commonly found in honey, salad dressing and shower gel dispensers. This valve deforms quite strongly when the bottle to which it is attached is squeezed (fig 1). The Spanish company Lékué manufactures kitchen utensils out of silicone rubber where the elasticity of silicone rubber is unconventionally exploited (fig 2). Products however, that are actually manufactured (partly) inside-out to make designs or manufacturing possible are very hard to find. Molding the product inside-out is exactly what makes the design of both the iValve versions possible.

The first product we made out of silicone rubber, which we used as a pilot project, had an inside-out feature designed into it: this feature limited the required number of parts while still fulfilling the requirements. The product is a stoma plug that enables the measurement of tracheostoma pressure during speech in laryngectomy patients. The feature that is molded inside-out functions as an airtight and comfortable cushion to the stoma (fig 3). This product has been successfully used in some studies.[83]

Economical impact and implementation

During the development of both the iValves and iPatches we kept in close contact with established tracheostoma device manufacturers. In order to develop an alternative or superior speech valve or adhesive patch it is crucial to analyze the already commercially available products. Our requirements are derived from those analyses and sometimes directly from suggestions from manufacturers.

For our devices to actually reach their users, these devices have to become available on the market. Within our current economic system, that means these products have to successfully compete with other products on the market. Therefore, our iValve has to be at least similar but preferably better than already available devices.

Since the most widely used automatic speech valve is the Atos FreeHands, this valve will be used here as our reference for marketability. The same goes for our iPatch adhesion patch: we will use the Atos Provox Regular adhesive patch and the Xtrabase for reference: since the Regular is the lowest priced patch, and the Xtrabase is the most widely used patch amongst FreeHands speech valve users.

There are a few financial considerations when it comes to market viability.

1. *Expanding an existing market:* If a company were to produce our valve as an alternative to their own products then this valve would address patients that had not been addressed by previous models.
2. *Reducing cost with follow-up product:* A company could decide to produce our valve, because it is more economical to produce than the previous product.
3. *Competing on a market:* a company could enter this market with our valve and compete with existing products: the total cost of ownership to the patient or imburser would then have to be lower or another clear benefit would make the financial trade-off beneficial (reduce cost elsewhere).

4. *Entering a new market:* our iValve can be used to address a new market previously unaddressed. If total cost of ownership can be kept low then the upcoming BRIC markets (Brazil, Russia, India and China) can be targeted, making speech restoration available to less wealthy patients.

Our valve will address patients that were previously not able to use an automatic speech valve. Because of its lower speech pressure, the valve can now be combined with relatively poor fitting adhesive. Patients who previously had difficulty using the Atos Provox FreeHands valve because of the problems associated with exhalation-operated valves may find the iValve a better alternative.
There are therefore numerous business-model possibilities to introduce the iValve on the market successfully, depending on the interested industrial partner(s). The business model could range from an introduction in developing countries, an introduction parallel to existing valves, or an introduction as a more economical valve – whether disposable or not – to compete with valves already on the market.

We can estimate the required total cost of ownership in a competing market, based upon the Atos Provox FreeHands valve. With an estimated lifetime of two years for the Provox FreeHands and a selling price of €715 excluding tax (price paid by insurance companies. Source: Atos Medical BV, The Netherlands) the price for the iValve will have to be priced similar or lower, depending on the valve's life time. The iValve with a one-year lifespan will have to cost €357 or less. A disposable single-use device will have to cost €1 per day. In the case of total single-use disposability with integrated filter, the price could increase to €5.70 (€1 + €4.70 for a comparable Atos Provox FreeHands Cassette filter). But that also depends on cleanability of the filter and how many filters patients use per day. If integration with an iPatch adhesive patch is considered, a price of €10.80 (€5.70 + €5.10 for a comparable Atos Provox Regular adhesive patch) or €15.40 (€5.70 + €9.70 for the Atos Xtrabase adhesive patch) per day would still be acceptable. It is clear that a considerable part of the total cost of handsfree speech is in the accompanying patches. If however we base our calculation on a half year use of the Atos FreeHands valve, the valve's price per day quadruples to a considerable €4, allowing for a maximum product per day price of around €18. The iValve's shot volume and weight are currently 3.92 ml and 4.5 gram (excluding runner). Industrial medical grade silicone rubber (used for injection molding of medical devices) costs €20 to €40 per kg (Wacker-Chemie Benelux B.V., Krommenie, The Netherlands) depending on the quantity. A single iValve would therefore cost around 1 to 2 €ct per valve in materials alone. Assembly (folding inside out), quality check and packaging adds to the piece price. Mold costs and handling equipment decrease with higher production volumes. A mold with the complexity required to mold the iValve v2 (an injection mold for liquid silicone rubber with one slider) will cost around €10k to €15k (source: Pezy Product Innovation, Groningen, The Netherlands) leading to piece prices including material of around €1.52 (10,000 pieces) or €0.17 (100,000 pieces), without handling/packaging. This price of course excludes all other costs such as marketing, development and overhead. Figures on the number of current laryngectomy patients are very hard to find since only statistics covering new cases of larynx cancer are registered. Dutch estimates

are between 2500 to 3000 patients (source: NSvG Patient Foundation, The Netherlands). Also, product pricing differs from country to country depending on use and imbursement system. The pricing of a novel product will thus have to be considered per country as well.
Our very low production costs however, make all of the above-mentioned business models possible.

Trends in healthcare

Prosthetic or orthotic devices are intertwined with surgery and medication. They append and compensate each other. Good surgery can make cumbersome prostheses obsolete, while smart and economical prostheses can make inherently risk-bearing surgery obsolete. All strive to restore the quality of life of the patient to a level similar to that when the patient was still healthy and normal. Regenerative medicine has been a promising field of medicine for many years and is said by many to be the alternative to prostheses and orthoses: why artificially compensate for the loss of a body function if you can regrow the function itself? Cancer treatment is also improving year by year, with early and better diagnosis and improved treatment and surgery. Notably the incidence of larynx cancer is on a steady decline in the Western world.[84] Will speech prostheses therefore eventually become obsolete? Although this might be the case in the future, it is currently not so. In the short term, we have patients who need prosthesis: there is an obvious need and market. But most importantly: while treatment and therapy are improving in the Western world, the majority of people live in less wealthy circumstances. And while the wealth in less privileged countries is increasing, patients who were previously unable to obtain treatment are now slowly gaining access. It has been demonstrated that low income, even in wealthy countries, limits access to the healthcare and therapy that are more economical in the short term. In the US this trend is visible in cancer therapy when laryngectomy is often preferred above the less-handicapping radio therapy and/or chemotherapy in lower household income families.[8] These patients are forced to give up speech due to economic reasons while the more benign, and speech capacity saving, treatments are comparably effective.[3] This means that the patient population needing the more economical prostheses will grow enormously whereas a relatively small percentage of patients will gain access to innovative, widely available but costly regenerative medicine. In that sense, the prosthetics market will simply shift.
Prosthesis and orthoses will have matured as well by that time. A foreseeable trend is that medical devices and our everyday devices will merge and that even the clear distinction between “medical” products and “health” products will disappear. Medical devices will become more personal and move toward being consumer market devices.

The other current trend is that of an ageing population. The often mentioned alarming rise in healthcare cost means business, research and innovation opportunity on the other side. This will be especially the case in the Netherlands, where the baby boom generation is considered to be the generally most wealthy and highly educated generation ever. This new generation will generate patients used to luxury and quality and willing to pay for that. A movement of patient liberation will be the result.
And this patient liberation is highly needed. It is surprising to find that patients endure ill-

designed medical solutions without as much as a complaint. The poorly fitting adhesive patches in this thesis are a good example. Amongst patients are engineers, craftsmen, tailors, generally handy people: yet only a very small number of patients rise to the challenge and use scissors and band aid to improve the offered patches (as shown, for instance, in chapter 7, fig. 1).

The next generation is not going to be so placid and might see medical technology as just another consumer good: people will demand quality and will pay for that. Currently people still refuse to pay for a speaking valve if it is not reimbursed by their health insurance. They perceive the product to be very expensive without reimbursement, although many are willing to pay the same amount for luxury goods. As demands on medical technology for quality and luxury will stimulate product diversity, the reimbursement from insurance companies will cease to be self-evident. Insurance companies will give patients a budget to choose from a range of products and they can pay extra for a preferred solution. Health insurance costs will be basic, extras can be chosen in freedom.

This new demand will also break the barrier between medical and consumer goods. Already the hand-held Nintendo DS game console can be equipped with a blood-sugar sensor, integrating gaming and therapy for improved therapy compliance. Wellness and health institutes will merge. Increased healthcare costs will become increased healthcare profit.

Our research continues. We are still looking for better solutions to improve lives and products. As of writing many projects are in different stages of development. This thesis shows only a few of those products.

Completing the circle

We can only evaluate the success of the products described in this thesis – the iValve v1 and v2 and iPatch v1 and v2 – once they are on the market and are used by patients. Feedback on how the iPatch and iValve are perceived by patients, how they are used, and how they fail is very important, and can be generated only during real use. A successful product, with a proper feedback process in operation at that stage, will help in the creation of a long and successful market presence.

For that last phase, commercial partners, with experience and connections in the market, are required. Only after successful commercialization can the product be said to have truly reached its goals. These outcomes should feed back into the design method and be used for redesigns and design improvements: eventually these products will become part of the "commercially available" product range mentioned throughout this thesis. Then one day our products will feature in the problem definition of other developers, working to improve the quality of life of laryngectomy patients even more. We wish them fun, luck and every success.

Summary

Summary

Patients who have undergone the surgical procedure of laryngectomy as a treatment for cancer in the neck have lost their voice box as a result of this procedure. They breathe through an artificial opening at the front of the neck: a tracheostoma. Laryngectomy can be a successful treatment against cancer, but leaves the patient mutilated: unable to speak and unable to breathe through their nose. Both speaking and air-conditioning of inhaled air (normally a function of the nose) can be restored using a range of commercially available devices. Speech is restored with a (routinely) inserted shunt prosthesis that connects the windpipe to the gullet, and air-conditioning is achieved by placing filters over the stoma. The patient is then required to close the stoma, usually with a finger, in order to produce speech. Although able to speak, a negative side effect is that the patient points at his/her handicap and always requires one hand when speaking. To avoid closing the stoma with a finger, automatic handsfree speech valves are available. Although these automatic speech valves, and also the stoma filters, seem to solve the afore-mentioned problems, these devices are actually not as widely used as one might think from the reported benefits to the patient. The reason is that both filters and automatic speech valves require an adhesive patch for their attachment to the stoma. The problem is that these adhesive patches do not last long, and that automatic handsfree speech valves are found to be tiresome and cumbersome in use.

Therefore, a voice restoration project was set up in which we used a design method to develop better devices. This method (explained in Chapter 1) entails first of all an extensive analysis of the problem, before the quest to find ways to reach the goals begins. In our problem analysis (Chapter 1) we found that only 15-25%[9-11] of patients uses automatic speech valves in particular and, in Chapter 2, we show that only 58% use an adhesive patch for attachment of any speech valve or filter, in general. In countries without a healthcare reimbursement system, only 9% of patients use an adhesive patch.

In Chapter 2 we also show that there are clear distinctions in the geometry of the stoma and surrounding tissue of the stoma (the peristoma) of patients; for instance, people who do not use patches have deeper stomas. Since most adhesive patches on the market are flat patches, it is obvious that most stomas are too deep to fit very well to these patches.

Currently the stoma is surgically altered, flattened, by incising the two sternocleidomastoid muscles along the stoma. In Chapter 3 we show that we could measure no difference between an incised and a non-incised stoma. In our opinion, it is better first to try to adapt the current adhesive patch technology to the patient, using the measurements from Chapter 2, before trying to adapt the patient's anatomy to fit the patch.

The reasons for low automatic speech valve use are the poor fit of adhesive patches, the loss of air in closing the valve, and not being able to pause mid-sentence without the valve opening again. These problems are due to the current closing principle of speech valves: automatic closure of the valve on strong exhalation. Due to this principle, exhaled air that is used during closing is not available for speech, and speaking pressure should to be maintained during speech to prevent the valve from opening again. Once these problems have been eliminated, 85.7% of patients interviewed in Chapter 6 reports that they prefer to use use an automatic speech valve.

After this extensive problem analysis we set out to increase the use of handsfree speech in laryngectomy patients by starting the development of an improved speech valve that is more likely to be used by patients. A new valve design was based upon the inhalation principle developed by Geertsema et al. In this principle the patient can durably switch the valve from breathing to speaking mode using strong inhalation, after which – and thus during speech – inhalation is still possible. The valve is switched back into breathing mode with a strong exhalation. Advantage of this valve principle is that speaking time can be extended, since inhalation is possible in mid-sentence, and no air is lost during closure. Chapter 4 describes the investigation of a mechanical principle that we used in the two speech valves: a functional bi-stable buckling rubber membrane that is used to durably switch from speaking to breathing. In Chapter 5, the iValve v1 automatic speech valve is tested for physiological characteristics: opening pressure (maximum speaking pressure), closing flow and air flow resistance. The valve worked as expected, except for the inhalation airflow resistance, which was higher than that of commercial valves.

In vivo (with real patients) evaluation of our prototype in Chapter 6 showed that patients preferred the iValve v1 over the commercial Atos Provox FreeHands valve and that speech quality was similar, with the added benefit that some patients were now also able to whisper using the iValve. The closing on inhalation instead of exhalation should lower the speaking pressure exerted on the valve and thus on adhesive patches: this could translate into a longer patch adherence time to the stoma. The greater size of the valve, the higher inhalation airflow resistance and some noise during inhalation were reported to be points that could be improved upon. Most importantly our prototype already performs comparable to that of the mature Atos FreeHands valve.

In Chapter 7, the recent developments section, we describe the iValve v2 version in which all the reported points (of Chapters 5 and 6) are addressed. The iValve v2 has a similar size to the Atos Provox FreeHands valve, and features the same low air flow resistance. Pilot patient testing of the improved prototype shows very promising results but a full patient study will have to be established for permanent proof.

The geometry data of Chapter 2 were used to address the issues of poor fit of devices to the stoma. Two contoured adhesive patches were developed: the concave iPatch v1 and the even more anatomically shaped concave iPatch v2. The v1 version is currently in full patient testing.

In conclusion: with respect to our goal to increase handsfree speech in patients, we succeeded in analyzing the problems of handsfree speech (ill-fitting adhesive patches, speaking pressure that is too high, and fatiguing speech valves) and we formulated the requirements for improved patch design and speech valve characteristics. Subsequently, we set out to improve the situation by developing a new handsfree speech valve. The iValve v1 already works just as well as the commercially available Atos FreeHands valve and is appreciated by patients. The shortcomings of the version 1 have all been addressed in v2, but a full *in vitro* (in the laboratory) and *in vivo* study will have to prove this. Adhesive patch design was based upon our measured geometry data and delivered the iPatch v1, which is currently under patient study, and the anatomically contoured iPatch v2, which will be tested in the near future.

We hope these devices to significantly contribute to a wider use of automatic speech valves, and easier, more natural and enjoyable speech. Only commercialization can proof whether we are right, and if indeed our new products are a success. If they are, they will become part of the "commercially available" products mentioned throughout this thesis. Then one day they may feature in the problem definition of other developers, working to improve the quality of life of laryngectomy patients even more. We wish them lots of success.

Samenvatting

Samenvatting

Bij patiënten die een laryngectomie hebben ondergaan, meestal ter behandeling van strottenhoofdkanker, is het strottenhoofd inclusief stembanden verwijderd. Na deze operatie ademen deze patiënten door een kunstmatige opening in de hals: een zogenoemd tracheostoma (letterlijk: luchtpijpmonding). Laryngectomie is een succesvolle, maar ontzettend ingrijpende behandeling tegen deze vorm van kanker. Na de ingreep kan de patiënt niet meer spreken en niet meer door de neus ademen. Gelukkig kan zowel het spraakvermogen als de luchtbehandeling door de neus (filteren en bevochtigen) met hulpmiddelen worden hersteld: het spraakvermogen door het inbrengen van een verbindingsventieltje tussen luchtpijp en slokdarm (al decennia een standaardprocedure), de luchtbehandelingfunctie van de neus door het plaatsen van filters voor het (tracheo)stoma. De patiënt kan nu weer spreken door het stoma met een vinger af te sluiten, Zo wordt de uitgeademde lucht door het verbindingsventieltje de slokdarm in geperst. De vibrerende slokdarm fungeert dan als stemband. Nadeel is dat de patiënt elke keer naar haar of zijn stoma wijst én altijd één hand nodig heeft om te kunnen spreken. Om dit te verhelpen zijn er automatische spreekklepjes ontworpen. Hoewel deze spreekklepjes in combinatie met stomafilters alle genoemde problemen lijken op te lossen, worden ze toch beduidend minder gebruikt dan men zou verwachten. De reden is dat het gebruik van automatische spreekklepjes als onhandig en vermoeiend wordt ervaren en dat zowel voor de klepjes als de filters, pleisters ter bevestiging rond het stoma gebruikt moeten worden. Deze pleisters blijven over het algemeen niet goed zitten.

Om dat te verhelpen hebben we een project voor spraakherstel opgezet. Met behulp van een ontwerpmethode hebben we getracht betere of in elk geval alternatieve spreekhulpmiddelen te ontwikkelen.

De toegepaste ontwerpmethodiek schrijft voor eerst op systematische wijze uitputtend het probleem te onderzoeken en pas daarna naar de oplossingen.

Tijdens de probleemanalyse (zie hoofdstuk 2) ontdekten we dat slechts 15 tot 25% van alle patiënten automatische spreekklepjes gebruikt, en maar 58% stomapleisters in combinatie met wat voor hulpmiddel (filters, klepjes) dan ook. In landen waar deze hulpmiddelen niet vergoed worden, past overigens maar 9% een stomapleister toe.

In de studie van hoofdstuk 2 hebben wij stoma's opgemeten. Hier tonen wij aan dat er duidelijke verbanden zijn tussen het pleistergebruik en de geometrie van het stoma en de huid rond het stoma (de peristoma): zo vonden wij bijvoorbeeld dat mensen die geen gebruik maken van stomapleisters over het algemeen een dieper stoma hebben. En omdat de meeste pleisters op de markt vlak zijn, ligt het voor de hand te veronderstellen dat de meeste stoma's te diep zijn om goed bij die pleisters te passen.

Soms worden stoma's chirurgisch aangepast en vlakker gemaakt door het insnijden van de spieren die aan weerszijde van het stoma liggen: de sternocleidomastoïdei.

In hoofdstuk 3 geven wij aan dat er geen verschil is gemeten tussen de diepte van stoma's die op deze manier zijn ingesneden en de stoma's die dat niet zijn. Het heeft dan ook onze voorkeur om eerst te proberen de bestaande pleistertechnologie aan de patiënt aan te passen op grond van de gegevens van hoofdstuk 2, alvorens te proberen de anatomie van de patiënt aan te passen aan de pleister.

De slechte pasvorm van de stomapleisters, het verlies van spreeklucht bij het sluiten van het klepje en het probleem dat men niet halverwege een zin kan pauzeren zonder dat het klepje weer opent, verklaren het tegenvallend gebruik van automatische spreekklepjes. Deze tekortkomingen vinden hun oorzaak in het sluitmechanisme van de huidige spreekklepjes: het automatisch sluiten van het klepje bij krachtige uitademing. Door dit principe verbruikt men een deel van de uitgeademde lucht voor het sluiten van het spreekklepje en die lucht is dus niet beschikbaar voor het spreken zelf. Bovendien moet er tijdens het spreken een continue ademdruk aangehouden worden om te voorkomen dat het klepje zich weer opent. Zouden deze tekortkomingen zijn verholpen dan zegt 85,7% van de geïnterviewde patiënten gebruik te gaan maken van zo'n automatisch spreekklepje (hoofdstuk 6).

Op grond van onze uitgebreide probleemanalyse besloten we het gebruik van automatische spreekklepjes onder laryngectomiepatiënten te vergroten door de ontwikkeling van een verbeterde versie.
Dit nieuwe klepje is gebaseerd op het *inhalatie*principe dat is ontwikkeld door Geertsema et al. Bij dit principe schakelt de patiënt door een krachtig *inademing* het spreekklepje in een blijvende spreekstand. Hierna, en dus tijdens het spreken, is inademen steeds mogelijk zonder dat het klepje uit de spreekstand schakelt. Het klepje wordt weer in de ademstand geschakeld door een krachtige uitademingstoot. Een voordeel van deze methode is dat het de spreektijd kan verlengen doordat midden in de zin, tussen de woorden in, gewoon kan worden ingeademd én er geen adem wordt verspild aan het schakelen van het klepje.
In hoofdstuk 4 beschrijven wij het onderzoek naar het mechanisch principe dat we gebruikten in de twee door ons ontwikkelde klepprototypes: een functioneel bi-stabiel knikkend rubbermembraan, dat schakelt tussen de spreek- en ademstand.
In hoofdstuk 5 testen we de eerste versie, de iValve v1 genoemd, met de fysiologische waarden die in het lichaam voorkomen: de openingsdruk (maximale spreekdruk), het sluitdebiet en de luchtstromingsweerstand. Het klepje functioneerde zoals gehoopt. Wel bleek dat de luchtstromingsweerstand bij inademing hoger was dan in de vergelijkbare commercieel verkrijgbare spreekklepjes.
In de *in vivo* (met patiënten)studie van het klepje, weergegeven in hoofdstuk 6, tonen wij aan dat patiënten de iValve boven de commercieel verkrijgbare Atos Provox FreeHands verkozen, dat de spraakkwaliteit vergelijkbaar was, en dat als bijkomend voordeel sommige patiënten nu konden fluisteren.
Door het sluiten van het klepje op inademing in plaats van op uitademing zou de spreekdruk op het klepje, en daarmee de druk op de stomapleister, lager moeten zijn. Als gevolg hiervan zou de stomapleister, en dus het spreekklepje, langer op het stoma moeten kunnen blijven zitten. De grootte van het spreekklepje, de hogere inhalatieweerstand en de bijgeluiden bij inhalatie waren genoemde verbeterpunten.
In hoofdstuk 7, "recent developments" of "recente ontwikkelingen", bespreken we het iValve v2 prototype, waarin alle verbeterpunten van de hoofdstukken 6 en 7 verwerkt zijn. De iValve v2 is in grootte vergelijkbaar met het Atos Provox FreeHands spreekklepje en heeft dezelfde, lage, inhalatieweerstand. Voorlopige patiënttesten van deze verbeterde versie zijn veelbelovend maar een volledige patiëntstudie zal dat pas eenduidig kunnen bewijzen.

Vervolgens zijn de nek- en stomageometriegegevens van hoofdstuk 2 gebruikt om het probleem van slecht passende stomapleisters aan te pakken. We hebben twee voorgevormde pleisters ontwikkeld: de trechtervormige iPatch versie 1 en de anatomisch gevormde iPatch versie 2. Het eerste prototype doorloopt momenteel een patiëntstudie.

Conclusie

Wat betreft ons doel, het vergroten van het gebruik van automatische spreekklepjes, kunnen we concluderen dat wij de oorzaken van het geringe gebruik aan het licht hebben gebracht. Het gebruik van automatische spreekklepjes is laag door passende pleisters, te hoge spreekdrukken en vermoeiende klepjes.

Daarnaast hebben wij een programma van eisen opgesteld voor (toekomstige) verbeterde pleisters en automatische spreekklepjes. Vervolgens hebben wij twee nieuwe automatische spreekklepprototypen ontwikkeld. De iValve v1 werkt inmiddels net zo goed als het commercieel verkrijgbare Atos FreeHands spreekklepje en wordt door patiënten erg gewaardeerd. Voor de tekortkomingen van deze eerste versie hebben we oplossingen gevonden en die toegepast in de tweede versie. Dit lijkt gelukt, maar of de tekortkomingen hiermee inderdaad zijn verholpen moet nog bevestigd worden in een volledige *in vitro*-(in het laboratorium) en *in vivo*-(met patiënten)studie.

De nieuwe ontwerpen van de stomapleisters zijn gebaseerd op onze stomageometriegegevens. Het eerste prototype, de iPatch v1, wordt in een patiëntstudie onderzocht. Een tweede iPatch versie, nog beter aangepast aan de anatomie, zal in de nabije toekomst worden getest.

Wij hopen dat al deze ontwerpen zullen bijdragen aan een algemener gebruik van automatische spreekklepjes, zodat patiënten in staat zijn eenvoudiger, natuurlijker en plezieriger te spreken. Echter pas na commercialisering zullen we zien of onze producten daadwerkelijk succesvol zijn. Als dat het geval is, dan zullen ze gaan behoren tot de “commercieel verkrijgbare” producten die door dit hele proefschrift heen worden genoemd. Misschien verschijnen ze dan in de probleemanalyse van andere ontwerpers die zich ten doel hebben gesteld het leven van laryngectomiepatiënten nog verder te verbeteren. We wensen hen heel veel succes toe!

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Appendix

Section 1: Appendix to Chapter 2

How to use the presented data in stoma patch design. Two examples.

The presented data can be used to design new stoma interfaces. Here are two examples of how to use the data.

The patient groups not yet addressed by stoma patches have relatively deeper and smaller stomas, smaller distances fold-to-fold, and more inclined stomas.
To target the group of current non-patch users, the following parameters could be used:

- using the largest P95 (95th percentile) stoma vertical and horizontal diameter (25 resp. 20 mm) will prevent obscuring the largest stomas,
- using a mean stoma depth for non-patch users of 18 mm and using the elasticity of both the skin and patch material to bridge the individual deviation from the value,
- addressing the inclination of the stoma (of 20 degrees leaning backward, non-patch users) by applying a superior stoma depth of 21 mm (Mean u non-patch users) and a inferior stoma depth d of 11 mm (median d non-patch users),
- applying the smaller P5 (5th percentile) distance stoma to chin, stoma to sternum, stoma to patients left and right (resp. 10, 10, 7, 8 mm) while,
- respecting the overall smaller distance fold-to-fold of 34 mm within this patient group (versus 44 mm in patch users).

To target the largest possible patient group (for instance P5 to P95, regardless of patch use) with a new glueable patch, the following guidelines emerge from this study:

- using the largest P95 stoma vertical and horizontal diameter (25 resp. 20 mm) will prevent obscuring the largest stomas,
- applying a mean superior and inferior stoma depth of respectively 13 and 12 mm and using the elasticity of both the skin and patch material will bridge the individual deviation from the selected value, and
- applying the smaller P5 distance stoma to chin, stoma to sternum, stoma to patients left and right (resp. 10, 10, 7, 8 mm) will fit most patients but obviously leaves very little area for applying glue.
- As the average stoma inclination is almost zero (3 degrees), the resulting funnel shaped patch can be almost circular symmetrical but with an elliptical (ellipticity of the stoma is 0.8) opening.

Section 2: Patent summary

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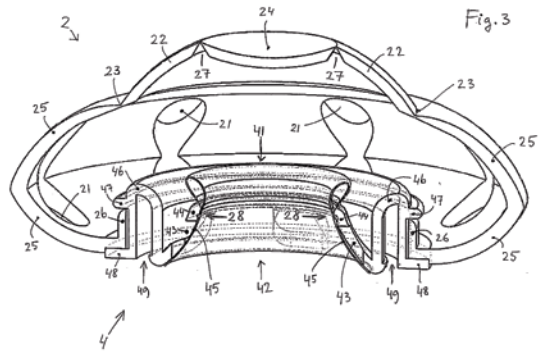

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Title **(EN)** TRACHEOSTOMA VALVE
(FR) VALVE DE TRACHÉOTOMIE

Abstract: **(EN)** A tracheostoma valve (1) comprises a main air passage structure (21, 41, 42) provided with a main valve arrangement. The tracheostoma valve is switchable in response to a spurt of inhalation air from a breathing condition into a speaking condition, while the tracheostoma valve is switchable in response to a spurt of exhalation air from said speaking condition into said breathing condition. The main valve arrangement comprises an at least partly flexible bistable wall portion (22, 23, 24) which has a circumferential bordering edge (23). At least part of the bordering edge forms a hinging means around which the bistable wall portion is hingeable between a first stable position in which an orifice (41) of the main air passage structure is being blocked, and a second stable position in which said orifice is released.
(FR) La présente invention concerne une valve de trachéotomie (1) qui comporte une structure de passage d'air principale (21, 41, 42) pourvue d'un agencement de valve principal. La valve de trachéotomie peut être commutée d'un état permettant la respiration en un état permettant l'élocution en réponse à une bouffée d'air inhalé, de même que la valve de trachéotomie peut être commutée dudit état permettant l'élocution audit état permettant la respiration en réponse à une bouffée d'air exhalé. L'agencement de valve principal comprend une partie de paroi bistable au moins partiellement souple (22, 23, 24) qui comporte un bord terminal circonférentiel (23).
Au moins une partie du bord terminal forme un moyen d'articulation autour duquel la partie de paroi bistable peut s'articuler entre une première position stable dans laquelle un orifice (41) de la structure de passage d'air principale est bloqué, et une seconde position stable dans laquelle ledit orifice est dégagé.

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The beauty about this job is that it helps people and that it is tremendous fun, and thus rewarding, doing it. Yet it can only be fun and rewarding if you do it with fun and rewarding people. This thesis is a feat of international and cross-disciplinary work; it is impossible to thank everyone that contributed to it. But I'll try!

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