Demands and properties of alloplastic implants for the treatment of stress urinary incontinence

Uwe Klinge†, Marcel Binneboesel, Stefanie Kuschel and Bernhard Schuessler

Surgical treatment of stress urinary incontinence changed dramatically with the introduction of the tension-free vaginal tape. Owing to its high efficacy and minimal patient discomfort this new minimally invasive procedure quickly obtained widespread acceptance and superseded the abdominal colposuspension as the gold standard. In the course of success of the original method a number of tension-free vaginal tapes flooded the market, varying in approach and material. These variations may strongly influence the safety, efficacy and long-term results of tension-free vaginal tape and its major modification, the transobturator technique. Therefore, it is the aim of this review to closely illuminate available materials and complications associated with this procedure. An extensive Medline search of the published literature up until 2006 on the subject of stress urinary incontinence was carried out. All sources identified were reviewed with particular attention to the method applied, the properties of the mesh materials and clinical complications. Apart from several technical variations, there are marked differences between the different vaginal sling materials, ranging from absorbable collagens over polypropylene to allografts. Although performed globally in substantial and increasing numbers, minimally invasive techniques for the surgical treatment of stress urinary incontinence are lacking sufficient safety data.

The concept of urethral slings for the treatment of urinary incontinence dates back to the beginning of the 20th Century [1]. The first modern urethral sling procedure was performed in 1942 by Aldridge, using rectus fascia [2]. In 1996, the surgical treatment of stress urinary incontinence (SUI) was dramatically altered with the introduction of the tension-free vaginal tape (TVT) by Ulmsten and Petros [3]. Its sustained success recently revived interest in different materials and surgical approaches that may potentially be used.

Since only a few countries have a national sling registry, valid information regarding the safety of these new materials and modifications is scarce. Therefore, it is the aim of this review to elucidate the necessity for the use of suburethral slings, understand the pathophysiological concept of SUI and mechanical demands of suburethral slings that aim to stabilize urethral hypermobility without causing obstructive voiding, and collect safety data for materials commonly used.

Concept of the therapy for SUI
SUI is defined by the International Continence Society as the complaint of involuntary leakage on effort or exertion, or on sneezing or coughing in the absence of any bladder contractions [4]. It affects one in every seven to nine middle-aged women [5] and imposes a significant burden on the women suffering from it [6].
The therapeutic options consist of behavioral intervention, physiotherapy or drugs for mild or moderate SUI and surgery for more severe cases or patients that have unsuccessfully tried the other options. The least invasive surgical therapeutic option is suburethral slings that follow the concept of restoring suburethral support. The procedure can be performed easily in an outpatient setting combining high-objective cure rates with minimal invasiveness to achieve improved quality of life for the patient [7]. The results were found to be equivalent to the former gold standard of abdominal colposuspension [8]. Whereas the classical TVT is inserted retroperitoneally, its major modification, the transoburator tape, avoids the retroperitoneal space by insertion through the transoburator foramen. The latter seems advantageous over the traditional method as it reduces the risk of inadvertent bladder and bowel injuries but inures a higher risk of nerve and vessel injury [9,10].

Pathophysiological background & mechanical demands
In order to define the biomechanical requirements of a vaginal suburethral sling, its principal action has to be clarified. An intrinsic sphincter deficiency with an increased relaxation of tissues forming the anterior vaginal wall and the endopelvic fascia are supposed to be the cause for the development of SUI [11]. Their reduced capability to work as thrust bearing for the urethra in case of elevated intra-abdominal pressure results in involuntary urinary leakage [12]. The underlying biochemical cause for the weakening of the tissues are still not fully understood and studies show controversial data on the collagen content of the endopelvic fascia around the urethra and skin of women with SUI; the majority, however, suggest it to be reduced [13–17].

By contrast, connective tissue samples of patients with pelvic organ prolapse (POP) had significantly higher scores of collagen [18] and a higher collagen type III expression was significantly correlated to the presence of POP [19]. However, it is conflicting that the collagen fibril diameter in both the SUI and POP subgroups was found to be significantly larger than that in the control groups [20]. Moreover, a significant increase in estrogen receptor concentration was seen in tissue probes of women with SUI or prolapse, despite reduced hydroxyl-proline content [21], as well as an increase in pro-matrix metalloproteinase (MMP)-2 expression, but inactivated MMP-2, MMP-9 and tissue inhibitor of MMP-2 [22].

A possible cause for the altered concentrations of collagen subtypes in women with SUI may be a disturbance of the translation of mRNA to collagen or by increased catabolism of collagen by collagenases [23]. However, micro-array analysis confirms that the alterations obtained in pelvic tissue from women with SUI are connected to many different genes or gene products. It could be shown that at least 62 genes involved in this process were up- and 28 were downregulated; it will be a challenging task to separate the causal change from the epiphenomenon [24].

It is still unclear whether the disturbed matrix composition indicates an impaired wound healing process, as could be shown for incisional hernia [25]. At least, the significantly reduced quantity of collagen types I and III in the pubocervical fascia of women with genuine stress incontinence III did not affect the efficacy of the TVT procedure up to 25 months of follow-up [26].

Initial biological suburethral vaginal slings have been replaced in favor of nonabsorbable synthetic materials but, independent of these modifications of the original method, high subjective and objective cure rates can be achieved [27]. As already implied by its name, the TVT should be placed suburethrally without any tension. Nevertheless, increased urethral resistance was described in two recently published papers that examined urodynamic changes following vaginal sling procedures [28] and it may be doubted whether or not many of the surgeons have the old bladder neck slings in mind that definitely aimed at least to form a backboard and functionally obstruct [29].

Apart from the integrity of the bladder neck and the urethral intrinsic sphincter mechanism, the intraurethral pressure is dependent on two factors: the surgeon adjusting the tape according to the intraoperative cough stress test that will be performed, under regional or local anesthesia, by filling up the bladder with 300–400 ml of saline and asking the patient to cough, in order to avoid retention tension strength should not exceed 3–4 kPa (0.3–0.4 N/cm²) [30]; and the mechanical properties of the vaginal sling itself.

The mechanical properties of different materials for TVTs differ widely. Whereas some products have a relatively high-tensile strength of 24 N [31], original TVT strips show a tensile strength of 7.5 N and biologic materials, such as cadaveric fascia lata, amounts to 4 N. Assuming an intraurethral pressure maximum of 8 kPa (0.8 N/cm²), an urethral diameter of 1 cm (0.8–1.5) and a band width of 1 cm, the tensile stress is calculated to be as high as 2 N or less, although the tensile strength decreases with time. For the TVT and the fascia lata, the tensile strengths accounted to 6 and 1.7 N after 6 weeks and 2.8 and 1.3 N after 12 weeks, respectively [32]. Therefore, long-term biologic fascia lata seems an unfavorable material for this indication.

Since the bladder filling up to 400 ml does not statistically affect the tensile stress weighing on the sling, and the sling only shows a slight movement during the Valsalva manoeuvre, there seems to be little need for the synthetic sling to be elastic [33]. Accordingly, an in vitro study has shown that a fascia strip could be stretched for a mean of 0.4 cm or less than 10% before sling tension began to increase [31]. It has been demonstrated that urethral and cervical mobility are unaffected by sling placement and that although the midurethra was immobile during episodes of stress, it appeared to be unaffected during voiding [12]. As could be visualized by dynamic perineal ultrasound, midurethral suspension surgery restricts the hypermobile bladder neck from opening the midurethral portion, providing increased resistance during stress to maintain continence [34]. Based on these data, the requirements of the ideal material for tension-free vaginal sling procedures should be a tensile strength of 2 N or greater, a width of 1 cm and little elasticity. Furthermore, the textile construction should have sufficient form stability to avoid any changes in width under the load.
Biomaterials & biological response

It has clearly been accepted that the remodeling of tissues around the vaginal sling and the tissue ingrowths into the sling result in significant differences comparing the pure textile properties with those of the sling-tissue compound. As known from various studies with meshes in hernia surgery, every implantation of synthetic materials is accompanied by a foreign body reaction. This foreign body reaction is characterized by the development of granulomas around the polymer filaments with accumulation of macrophages, increased apoptosis and proliferation corresponding with accelerated cell turnover, and marked expression of MMP-2 as an indicator of the remodeling of the extracellular matrix. In general, the activity of the inflammatory response depends mainly on the material and the structure used. Strictly correlating to the extent of inflammation there is a fibrotic reaction, embedding the implant into a scar field. Unfortunately, as with every wound, this scar tends to contract over time, which may reduce the area of the sling. Small pore polypropylene mesh constructions showed shrinkage of 20–80% [35–37]. Accordingly, transrectal sonography revealed that a 20-mm sling decreased to 13.5 ± 3.3 mm at 3 months [38]. Furthermore, a postoperative increase of the forces onto the urethra with overzealous sling tensioning may develop. This may result in an increased risk for obstruction in the case of materials with intense inflammation and fibrosis in biological and synthetic materials [39]. Correspondingly, it is reasonable to believe that simple loosening of the sling tends to fail as the subsequent wound healing process will soon reconstitute the previous scarry sponge.

Biological tissue slings

Biological tissue slings are generally well tolerated. Autologous material for pubovaginal slings have been taken from rectus fascia or fascia lata. Alternatively, human allograft cadaveric fascia or dermis, animal xenograft dermis, or intestinal submucosa have frequently been employed as alternative sources for sling material as they do not require an additional procedure with potentially longer recovery times. Xenografts may potentially carry the risk of viruses and DNA transmission [40]. The implanted matrix of biological implants mainly consists of collagen that have been more or less denaturated during the process of manufacturing to preserve them and obtain standardized samples. However, in the long term, biological implants will be degraded and replaced by scar tissue [41]. When reoperating on patients who had received cross-linked porcine dermis grafts, the implants appeared to be completely replaced by dense fibroconnective tissue with variable tissue reactions that may have unpredictable clinical outcomes in different patients [42]. This raises the question regarding the overall tolerability and efficacy of biological implants in pelvic reconstructive surgery [43]. Collagen materials pretreated with an increased concentration of formaldehyde and glutaraldehyde demonstrated increased cross-linkage with delayed absorption time, whereas nonmodified autologous materials were usually replaced within 6–12 weeks [43]. When undergoing an inflammatory foreign body reaction and collagen remodeling, the highly cross-linked materials may retain their original size for several weeks [44–46]. However, as could be shown with two porcine-derived biomaterials, the extended scarry reaction may not prevent the shrinkage process [47] but may favor outlet obstruction [48].

In a recent review by Bukkapatnam and Rodriguez [49], slings deriving from human or animal donors were found to be degraded and, therefore, to be less durable than synthetic slings [12]. Accordingly, all collagen-based materials have to be regarded as at least absorbable in the long term, which might be in conflict with the assumption that the basic disease is an ongoing weakening of the tissue. This might be the reason that there are some disappointing results, for example, with acellular cross-linked porcine dermis [50–52]. However, other reports have demonstrated that autologous grafts used in pubovaginal slings are superior to cadaveric fascia [53–55] and even better than synthetic slings [56]. The differences in evaluating these materials may also depend on the length of the follow-up period. Owing to the aforementioned graft degeneration long-term results may be worse [57,58]. Furthermore, the different results may be caused by the widely unknown details of manufacturing biological grafts leading to materials with differing structures and collagen properties that are difficult to compare. Standardization and lack of degeneration are the most important factors for the increased use of synthetic slings.

Alloplastic tissue slings or synthetic slings?

Synthetic slings are mostly composed of polypropylene (PP), silicone, polytetrafluoroethylene (PTFE), polyvinylidene fluoride (PVDF) or polyester.

PP is the most popular utilized synthetic material today. It is a thermoplastic propene material with a melting point at 160°C and a molecular weight of 100 kD. The filaments attain the strength of steel, while its density is only an eighth of that of iron. Until recently, complete degradation for polyester has not been described. Nevertheless, measuring the yarn tension of PP-suture material after 7 years, a tension decrease of 50% under hydrolytic condition was observed. PP meshes are usually made of monofilaments that make them comparatively stiff and inelastic. Pubovaginal slings are produced from strips of PP mesh-hosery and, therefore, often reveal sharp-cut filaments that once brought into place also have some self-anchoring properties (FIGURES 1 & 2). In order to be able to pull the sling through the tissue they are wrapped in a plastic cover, which is removed after final placement of the sling.

PP filaments are usually thick and cause an intense inflammatory response with lots of macrophages. After 1 week there is increased fibrotic encapsulation with stimulated remodeling at the interface, indicated by increased levels of apoptosis and proliferation [59–61]. It is still unclear whether single-ending fibers stimulate this chronic inflammatory process but abdominal adhesions often found at the border of meshes placed intraperitoneally support this. The intense scar formation contributes to the wound contraction that is found to be approximately 20–30% [62].
Through quick-reacting industry there are a variety of PP midurethral slings available, differing in many small but potentially important details from the original monofilament TVT (Ethicon® Inc., NY, USA), primarily described by Ulmsten in 1996 [63], the monofilament SPARC™ Sling System (American Medical Systems, MN, USA) or the multifilament intravaginal slingplasty (IVS; Tyco Healthcare®) [64]. Both monofilament slings are placed with minimal tension beneath the midurethral complex. They are enwrapped by a protecting plastic sheath to prevent mechanical injury due to the rough side with sharp leaving fibers. The sheath is removed after the sling has been placed at the correct site. The SPARC sling includes a knotted tensioning suture that runs longitudinally through the sling, which was designed to decrease the stretch on the mesh during intraoperative plastic sheath removal and also during possible later readjustment. Contrary to porcine small intestine submucosa (Surgisis®) Marlex®, another PP mesh, demonstrated a more pronounced inflammatory response, although after 90 days both materials were enwrapped by a similar intense fibrosis [65]. Although PP mesh, cadaveric fascia and porcine dermis all induce a marked inflammatory and fibrotic reaction, the highest amount of scar tissue is formed around the synthetic material [66]. In comparison with the IVS, the TVT demonstrated an increased inflammatory reaction, whereas the amount of collagens did not differ [67].

Polyester is a macromolecular compound. Similar meshes made of polyethylene terephthalate result from polycondensation of carbonic acids. The texture of polyester prostheses is predominantly multifilament, resulting in high flexibility. The initial tissue reaction demonstrates a moderate inflammation. The major disadvantage of polyesters are their obligate degradation via hydrolysis. After several years, polyester grafts lose most of their mechanical stability [68,69]. This process may even be accelerated in the presence of a smoldering infection [70]. In 1998, Leber and colleagues found significantly more infections, recurrences and fistula after use of polyester meshes for incisional hernia repair and, therefore, concluded that the mesh should not be used for hernia repair [71].

Recently, reports on transvaginal excisions of Dacron bolster [72] and urethral erosions of TVT, and woven polyester pubovaginal sling have appeared in the literature [73]. Compared with fascia lata, wound infection rates were twice as high for polyethylene slings [74]. Another polyester sling coated with bovine collagen (ProteGen sling; Boston Scientific, MA, USA) was even withdrawn from the market in 1999 because of its increased risk of fistula formation [75,76].

PVDF is probably the synthetic material with the best biocompatibility, minimal foreign body reaction and optimal ingrowth (FIGURE 3) [83]. PVDF is a nonabsorbable fluoropolymer consisting of alternating methylene and difluoromethylene groups, and demonstrates improved textile and biological properties compared with other materials. It is thermally stable and shows an exceptional chemical stability and excellent resistance to aging with preserved mechanical stability, even after long-term incorporation [84]. PVDF sutures are routinely used in cardiovascular and orthopedic surgery. It has shown a minimal inflammatory and fibrotic foreign body reaction. Therefore, it combines the minimal cellular activation of PTFE and the capability for the construction of porous meshes as with PP. However, until now, clinical experience for this material in surgery for female SUI is limited.

Further materials that have been used are silicone or polyglactin in combination with fascia [85]. The latter demonstrated favorable preliminary results. Recent studies have found an unproportionally higher risk of erosive phenomena with slings...
and meshes used for sacrocolpopexy that partly or entirely consist of silicone ranging from 8.8–71% [86-90]. Therefore, silicone appears to be an inappropriate material for this indication.

**Tissue integration & foreign body reaction**

In contrast to film-like materials and silicone that form a fibrotic capsule, porous structures show an ingrowth of tissue in between the filaments. Contrary to early suggestions, the structure of the device is even more important for the intensity of the foreign body reaction than the polymer itself [91,92]. It depends largely on the pore size and whether they are either filled with scar tissue (<1000 µm) or with fat tissue (>1000 µm). In the case of small pores, the inflammatory and fibrotic reaction entirely bridges the pores resulting in scar plates with the mesh as core. In particular, small pore constructions tend to show an intense remodeling at the surface with excessive fibrotic capsule formation. The persistent cell-turn-over may favor migration and fistula formation. The degree of difficulty in removing the synthetic material correlates with the amount of time of tissue ingrowth, which already takes places within the first 2 weeks. This may explain why film-like materials, such as PTFE and silicone, which will not grow through, within the first 2 weeks. This may explain why film-like materials, such as PTFE and silicone, which will not grow through, may loosen and therefore lead to the recurrence of SUI [90,93].

Although it is well known that the activity of inflammation highly correlates with the quantity of scar formation, there are few investigations regarding the quality of the scar tissue induced. Analyzing the collagen surrounding hernia meshes, Junge and colleagues found an impairment of the quality of the extracellular matrix deposition as determined by collagen type I/III ratio in contrast to scars without the presence of synthetics [94].

**Complications**

According to the current literature the overall morbidity of tension-free vaginal slings amounts to 12.5% [95]. As for most other solid implants, infection and migration are the most serious adverse events that usually manifest with a considerable delay of several months or even years [12]. Infections are usually caused by an intraoperative colonization of biofilm-forming bacteria that may reactivate in the patients’ later life and present as late abscesses or fistula. Even though the sterile handling of the implant and the intraoperative application of intravenous antibiotics significantly reduces infection rates, late-onset infections are mainly dependent on the patients’ individual immune system.

Migration is a phenomenon that it remains unclear how to prevent for certain. It is controversially discussed, whether fixing to the pubic bone/tubercle by bone anchors should be avoided, due to some cases of severe osteomyelitis that may be resistant to common therapies [96]. The implant may even migrate through tissues, such as the urethra, or move towards the bladder or vagina [97]. Vaginal erosions continue to be a common complication of the TVT procedure with an incidence of 10–14% despite limited follow-up of 2 years or less [98] and are probably underestimated in some other studies with 1.1% [99]. Reduction of the tension on mesh materials with large interstices seems to reduce erosion/protrusion rates [28]. Urethral erosions were more common early in the transobturator approach experience and are thought to be related to the silicone coating used in the midportion of some of the original transobturator slings. It is interesting to know that up to 50% of vaginal erosions are asymptomatic whereas urethral erosions have been reported to be mainly symptomatic [100,101].

In the largest study to date on the transobturator urethral sling approach for SUI, Boccon-Gibod and colleagues reported results of 441 patients [111]. Tape extrusion occurred in 13 out of 233 patients using silicone-coated nonwoven Uratape® (5.6%) and in four out of 208 patients using nonwoven ObTape™ (1.9%). Since the application of a small pore, PP meshes used in prolapse repair have demonstrated unfavorable results, with a vaginal erosion rate of 20% [102]. The authors strictly recommend the use of large-pore material meshes/slings as already applied and accepted in surgery for hernia [103]. Recently, Meschia and colleagues analyzed the 2-year outcome of TVT and IVS. They found a cure rate of 87 and 78%, respectively, but also a 9% vaginal erosion/infection rate in the IVS group requiring removal, compared with none in the TVT group [104].

A high intensity of local tissue remodeling in combination with shear forces may favor dislocations. Dislocation of the sling into the bladder was supposed to be the cause of persistent SUI in some clinical reports [105].

However, one has to take into account that fistula formation has also been described with the use of nonsynthetic materials, acellular cross-linked porcine derrmis [106], dermal allografts [107] and even with autologous rectus fascia pubovaginal sling [108,109].

**Conclusion**

The complication rate with the use of an alloplastic sling/mesh is difficult to estimate. In particular, the rate of long-term complications seems to deserve more attention [110-113]. However, recent studies and honest reports on complications help to profile the ideal sling material, which has yet to be found [114]. Synthetic materials have the advantage of being highly standardized.
in their biomechanical properties and their induction of a foreign body reaction, but their success and complication rate widely depends on the kind and amount of material; in addition, long-term reliance has to be warranted. PP is the material used most often but, in consideration of experimental data, PVDF may be a promising new material. Due to an excessively high rate of erosion, silicone should not be used. The requirements for synthetic suburethral vaginal tension-free slings are listed in Box 1. According to surgical hernia repair the use of large-pore and weight-reduced materials is mandatory. In addition, new devices for pelvic surgery with improved biocompatibility may prove their superiority even in long-term follow-ups and may negate the conclusions of past meta-analyses, which suggested using biological instead of synthetic materials [12]. Possible complications related to vaginal slings should be explained prior to obtaining preoperative informed consent. Since some complications present with a considerable delay, careful clinical examination follow-up appears to be indispensable.

Expert commentary
Conventional suture techniques for the treatment of genital prolapse and SUI encumbered with recurrence rates of 20–30% have been replaced by biological or synthetic bands, or meshes, to lift the insufficient pelvic floor. This approach considers a weakness of the tissues and provides a scary strengthening, which should last the patient's lifetime. The best materials seem to be nonabsorbable alloplastic materials as they show standardized and reproducible physicochemical characteristics. However, the use of these implants has to take into account the obligate foreign body reaction with chronic inflammation at the interface with risk of migration, erosion and infection. Erosion is mainly seen with rigid, low-porosity, thick meshes, whereas film-like or multifilament materials may be prone to infection.

Wound contraction of the fibrous integration leads to shrinkage of the implant area with functional worsening. As experienced in other medical fields with net-like implants, in particular surgery that uses meshes to reinforce a hernia repair, these complications are closely related to the porous structure of the implant. In the future mechanical requirements for the pelvic floor will have to be defined more precisely in order to design devices with optimal functional and biological properties.

Five-year view
The treatment of pelvic floor insufficiency has passed the period proving that porous textile structures are superior to former suture techniques. In the future, techniques even for complex prolapses will be optimized. Indications have to be

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**Box 1. Proposal of requirements for pubovaginal slings**

- Large pore (>1 mm, no film) mesh sling
- Little elasticity (<10%)
- Width of 1 cm or more as a smaller size would encourage cutting through the urethra (causing stability under load)
- Little inflammation and fibrosis with little cell turnover at the interface and little shrinkage
- No sharp leaving fibers as they may stimulate an inflammatory response
- Tensile strength not necessarily more than 2 N/cm; limited tension to the sling of 3–5 kPa, hopefully objectified by future intraoperative measurements
- Nonabsorbable, permanent, synthetic, instead of collagen, devices

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**Key issues**

- Modern techniques for the treatment of urinary incontinence include the use of urethral slings and tension-free vaginal tapes.
- Reinforcement of local tissues by implants is able to compensate for intrinsic sphincter deficiency, as well as to strengthen the relaxed pelvic floor.
- Biological devices mainly consist of collagens, which demonstrate considerable variations in strength and absorption depending on the cross-linking. When the device is replaced by scar tissue recurrence may occur, in particular in patients for whom a defective collagen metabolism is discussed.
- Nonabsorbable, synthetic devices made of polypropylene, polyester, polytetrafluorethylene or polyvinylidene fluoride, are more standardizable but have the disadvantage of an obligatory chronic inflammatory foreign body reaction, which, over time, may result in migration and erosion.
- Large-pore constructions (>1000 µm) have a reduced intensity of the inflammatory and fibrous tissue response and, therefore, may reveal an improved biocompatibility.
- An important consequence of the dense scar tissue induced by some synthetic devices is the shrinkage of the wound area with consecutive obstruction of the urethral flow.
- Despite encouraging early results, some reports regarding the high rate of infection, pain or migration of the device discourages a routine use of synthetic implants. A long-term survey will be necessary to identify the patients where the functional success justifies the risk for delayed complications.
clarified to select the least invasive access to place the mesh, perineally, transvaginally, transabdominally (laparoscopic or open) or a combination of these. The shape of the device has to be adjusted to the functional demands, as a sling or flat mesh, or a combination of both. Porosity and textile structure has to consider the mechanical elasticity and strength needed without permitting lateral enrolment. Devices will be developed that induce less periprosthetic inflammation and fibrosis but provide durable and resilient reinforcement of the pelvic floor.

Analysis of the patients’ tissue and genes will help to understand the basic defect of the disease and thereby may offer the possibility to improve tissue restitution by local drug administration.

References
Papers of special note have been highlighted as:
• of interest
  • of considerable interest

A prospective multicenter randomized trial of tension-free vaginal tape (TVT) and colposuspension for primary urodynamic stress incontinence with 2-year follow-up found 63% of the TVT group and 51% of the colposuspension group to be objectively cured at 2 years.

- Experience of TVT and transobturator tape procedures in 440 patients.
  
24 No difference in the amount of mRNA of procollagen type I and III but of the quantity of collagen type I and III in patients with stress urinary incontinence (SUI).
Tape position varied from 30 mm above
and between 15 mm above and 18.7 mm
below the symphysis at rest.

Reduced quantity of collagen types I and
III in the pubocervical fascia does not affect
the efficacy of the TVT procedure.

Urogenital tract erosion and pelvic abscess
are significant complications of suburethral
tape.

Bladder volume did not statistically affect
the sustained tension, which is situ in far
less than the maximal load needed to
break fascial strips.

Tape position varied from 30 mm above
to 12.7 mm below the symphysis at rest
and between 15 mm above and 18.7 mm
below the symphysis onValsalva.
A total of 34 women who underwent removal of a polyester sling secondary to erosion, infection or pain at a mean of 7.95 months (range 1 to 22) after sling placement.

Amid P. Classification of biomaterials and their related complications in abdominal wall hernia surgery. Hernia 1, 5-8 (1997).

First attempt to classify different synthetic materials with regard to porosity.


Plate-like, dense synthetic material tends to cause wound infection.


Comparison of three polypropylene slings with various pores in the subcutaneous tissue of rats.


Comparison of various biologic and synthetic materials in an animal model with regard to inflammation and fibrosis.


Comparison of two polypropylene materials in a rat model.


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Amid P. Classification of biomaterials and their related complications in abdominal wall hernia surgery. Hernia 1, 5-8 (1997).

First attempt to classify different synthetic materials with regard to porosity.


High failure rate using fresh frozen cadaveric fascia lata.


A total of 37.6% of patients had recurrent moderate-to-severe (grades 2 to 3) SUI at follow-up.


Comparison of three polypropylene slings with various pores in the subcutaneous tissue of rats.


Silicone-coated polyester mesh is associated with a high rate of vaginal erosion when used as a transvaginal suburethral sling.


An acceptable rate of erosions led to the discontinuation of ObTape™.


Second sling in case of recurrence possible.


Overall complication rate following surgery for SUI in the USA was 13.0% (with bleeding, surgical injury, urinary/renal, infection, wound problem, pulmonary insufficiency, myocardial infarction and thromboembolism).


Bone anchors may lead to untreatable complications, such as osteomyelitis.


Described persistent painful or irritative symptoms after pubovaginal sling placement.


A total of 9% of women treated with the intravaginal sling plasty required removal of the tape for erosions.


Erosions can also occur with fascial slings.


Owing to a lack of well-designed prospective randomized trials, recommendations for using graft materials in vaginal reconstructive surgery should be limited to carefully selected patient populations.
Treatment of stress urinary incontinence


• Lack of prospective trials.


• Lack of prospective trials.


Affiliations

• Uwe Klinge, M.D
  Applied Medical Engineering, Helmholtz Institute, RWTH Aachen University, Pauwelstraße 20, 52074 Aachen, Germany
  Tel.: +49 241 808 9352
  Fax: +49 241 808 2442
  klinge@hia.rwth-aachen.de

• Marcel Binneboesel, M.D
  Surgical Department, University Hospital of the RWTH Aachen, Pauwelstraße 30, 52074 Aachen, Germany
  mbinneboesel@ukaachen.de

• Stefanie Kuschel, M.D
  Department of Gynecology and Obstetrics, Cantonal Hospital Lucerne, Spitalstrasse, 6000 Lucerne, Switzerland
  stefanie.kuschel@kld.ch

• Bernhard Schuessler, M.D
  Department of Gynecology and Obstetrics, Cantonal Hospital Lucerne, Lucerne, Switzerland
  Tel.: +41 412 053 501
  Fax: +41 412 055 932
  bernhard.schuessler@kld.ch